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Tradeoffs between image quality and dose

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Abstract Image quality takes on different perspectives and meanings when associated with the concept of as low as reasonably achievable (ALARA), which is chiefly focused on radiation dose delivered as a result of a medical imaging procedure. ALARA is important because of the increased radiosensitivity of children to ionizing radiation and the desire to keep the radiation dose low. By the same token, however, image quality is also important because of the need to provide the necessary information in a radiograph in order to make an accurate diagnosis. Thus, there are tradeoffs to be considered between image quality and radiation dose, which is the main topic of this article. ALARA does not necessarily mean the lowest radiation dose, nor, when implemented, does it result in the least desirable radiographic images. With the recent widespread implementation of digital radiographic detectors and displays, a new level of flexibility and complexity confronts the technologist, physicist, and radiologist in optimizing the pediatric radiography exam. This is due to the separation of the acquisition, display, and archiving events that were previously combined by the screen-film detector, which allows for compensation for under- and overexposures, image processing, and on-line image manipulation. As explained in the article, different concepts must be introduced for a better understanding of the tradeoffs encountered when dealing with digital radiography and ALARA. In addition, there are many instances during the image acquisition/display/interpretation process in which image quality and associated dose can be compromised. This requires continuous diligence to quality control and feedback mechanisms to verify that the goals of image quality, dose and ALARA are achieved.

Keywords Radiation dose · ALARA · Image quality · Digital radiography · Radiographic techniques · Image evaluation

Introduction

Pediatric radiography is a challenging procedure from the perspective of image quality and radiation dose. Certainly, radiation dose is an extremely important issue to the young child, who is significantly more radiosensitive and more likely to manifest radiation-induced changes over its lifetime. Children are approximately ten times more sensitive to radiation-induced cancer than middle-aged adults and three times more sensitive than the population average [1]. At the same time, the need for high image quality is absolutely necessary in order for the diagnostic benefits of the exam to be achieved.

This which often translates into more radiation exposure and higher dose. Digital imaging technologies in some cases are providing more x-ray absorption and conversion efficiency compared to conventional screen film, and have the potential to reduce radiation dose for a given image quality; but first we need to define what image quality is, and how it is achieved.

Image quality is an indicator of the relevance of the information presented in the image to the task we seek to accomplish using the image, and is evaluted in terms of portrayal of normal anatomy or the depiction of potential pathology. Image quality is not necessarily the same in all cases, but is an exam specific characteristic that is dependent on the image processing and contrast enhancement necessary to optimize the interaction of the human viewer and the information content portrayed in the image to render an accurate interpretation.

Implementation of digital radiography has changed the rules for the determination and achievement of image quality. Generally speaking, there are two classes of digital detectors, one emulating the screen-film paradigm, known as computed radiography (CR), and a multitude of detectors collectively known as direct radiography (DR) devices. While the nomenclature for these devices is undergoing revisions, the outcome of the acquired digital images is very similar. Unlike screen-film radiography where the film serves as the acquisition, display, and archive medium, and must be optimized during the acquisition phase of the image. The image quality is built into the properly exposed and processed film radiograph. Digital radiography separates these three components of the imaging process. This separability has potentially good and bad outcomes when using the imaging system. Optimization requires consideration of radiographic technique (kV, mAs, beam filtration), detector efficiency (composition, x-ray absorption, signal conversion), image pre- and post-processing (flat-fielding, dynamic range compression, digitization, contrast enhancement, spatial frequency enhancement), image display (film, soft copy, calibration, viewing conditions), and environmental/human status (e.g., alertness, fatigue of the radiologist).

The good aspects of digital separability allow for image processing, quantitative data extraction, application of computer aided diagnosis, and ability to use a variable-exposure detector for specific examinations. For example, a pediatric scoliosis exam might benefit from a low-dose (high-speed) acquisition, because diagnosis/evaluation is chiefly concerned with the high-contrast bony structure and geometry of the vertebral column (at least in a large number of sequential exams on the same patient). This allows a significantly reduced dose because the signal (anatomy of interest) is much greater than the background stochastic (mottle) and structure (anatomic) noise, as shown in Fig. 1.

The bad outcomes of separability are that there is no direct feedback to the technologist in terms of proper radiation exposure, technique adjustments are sometimes counterintuitive for repeat radiographs, and compensation of digital image presentation for bad techniques/exposure can lead to suboptimal use and complacency by the technologist. In short, a digital system requires more thought to use properly, and can easily be misused. From the perspective of dose management, it is easy to fall into the dose creep situation, whereby escalation of doses occurs because the images look good (noise-free) and one technique can be used for all pediatric examinations.... certainly NOT the intention of ALARA.

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A systems approach is taken to describe the various considerations that are important in optimizing the image (e.g., getting the best image quality necessary for the lowest possible dose). Overall image quality is based upon image fidelity, which measures the departure of a processed image from some standard image, and image intelligibility, which denotes the ability of human or machine to extract relevant clinical information [2]. Image fidelity and intelligibility are dependent on subject contrast, image resolution, image noise, signal-to-noise ratio (SNR), and contrast-to-noise ratio (CNR). Image pre- and postprocessing of the raw digital image is absolutely necessary in order to render an intelligible image. The extent that image processing can optimize the relevant information is determined by the image SNR and CNR. The amount of radiation dose needed to achieve an optimal SNR/CNR is dependent on the acquisition techniques (kVp and mA) as well as the efficiency of the detector in transferring the information from the acquisition event to the display event. One important measure of this information-transfer efficiency is termed the DQE (detective quantum efficiency), as explained in this review.

X-ray detection efficiency and the signal detection task: tradeoff between absorption efficiency and spatial resolution for digital x-ray detector systems

Efficient detection of the spatially dependent and intensity-modulated x-ray beam transmitted through the patient is the first and most important task of the x-ray detector. Ideally, 100% quantum detection efficiency is preferred. Materials used for x-ray detection are broadly

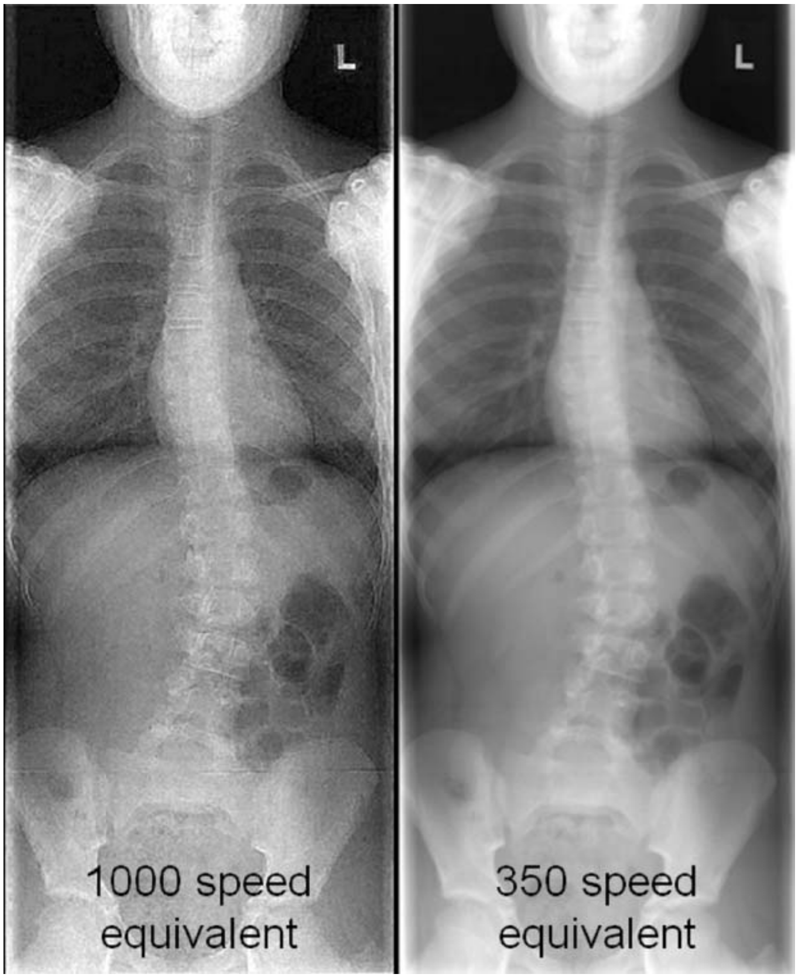

Fig. 1 An example of pediatric image quality at low dose (*left*) and high dose (*right*), achievable with digital radiography. It can be argued, according to ALARA principles, that both images represent good practice dependent on the needed fidelity to make the requested diagnosis

classified in three types: turbid, unstructured scintillators (e.g., Gd_2O_2S, gadolinium oxysulfide; BaFBr, barium fluorobromide), structured phosphors (e.g., CsI, cesium iodide), and semiconductor direct detectors (e.g., *a*-Se, amorphous selenium). For all x-ray converters, x-ray absorption is improved with larger thicknesses. In the case of unstructured phosphors, the thicker phosphor causes a loss of spatial resolution due to the diffusion of the light within the phosphor substrate; higher spatial resolution transfer requires a thinner scintillator, with a concurrent loss of detection efficiency. Structured phosphors can be made relatively thick without a corresponding loss of resolution, due to the light-channeling properties of the crystalline needle formations and the internal reflection of light that preserves the resolution, as shown in Fig. 2. Direct semiconductor detectors produce signal directly, and acquire the charge pairs under a large voltage, which minimizes the spread of the produced signal independent of thickness. Thus, systems with structured phosphors or semiconductor detectors can achieve good spatial resolution and high detection efficiency, as shown in Fig. 2.

However, the converted signals must also be efficiently collected. It makes no sense to have high detection efficiency without concomitant high conversion efficiency. Cases in point: (1) a thick, lead detector can achieve 100% detection efficiency, but cannot transfer the x-ray carrier information to a useful visible signal; (2) an optically coupled scintillator detector (e.g., scintillator-lens-CCD camera) system might have reasonable x-ray to light conversion efficiency, but poor lens collection efficiency, particularly at low doses, which produces output signals with less statistical integrity than those of the x-rays absorbed in the scintillator (this is known as a secondary quantum sink). Information transfer is thus dependent not only on the detection efficiency, but also on the collection efficiency of the information carriers, the conversion efficiency into a useful signal, and the spatial resolution transfer characteristics of the detector. A convenient measure of the latter attribute is termed the modulation transfer function (MTF), which describes the normalized object (signal) transfer characteristics in terms of signal modulation (loss) as a function of spatial frequency. Object

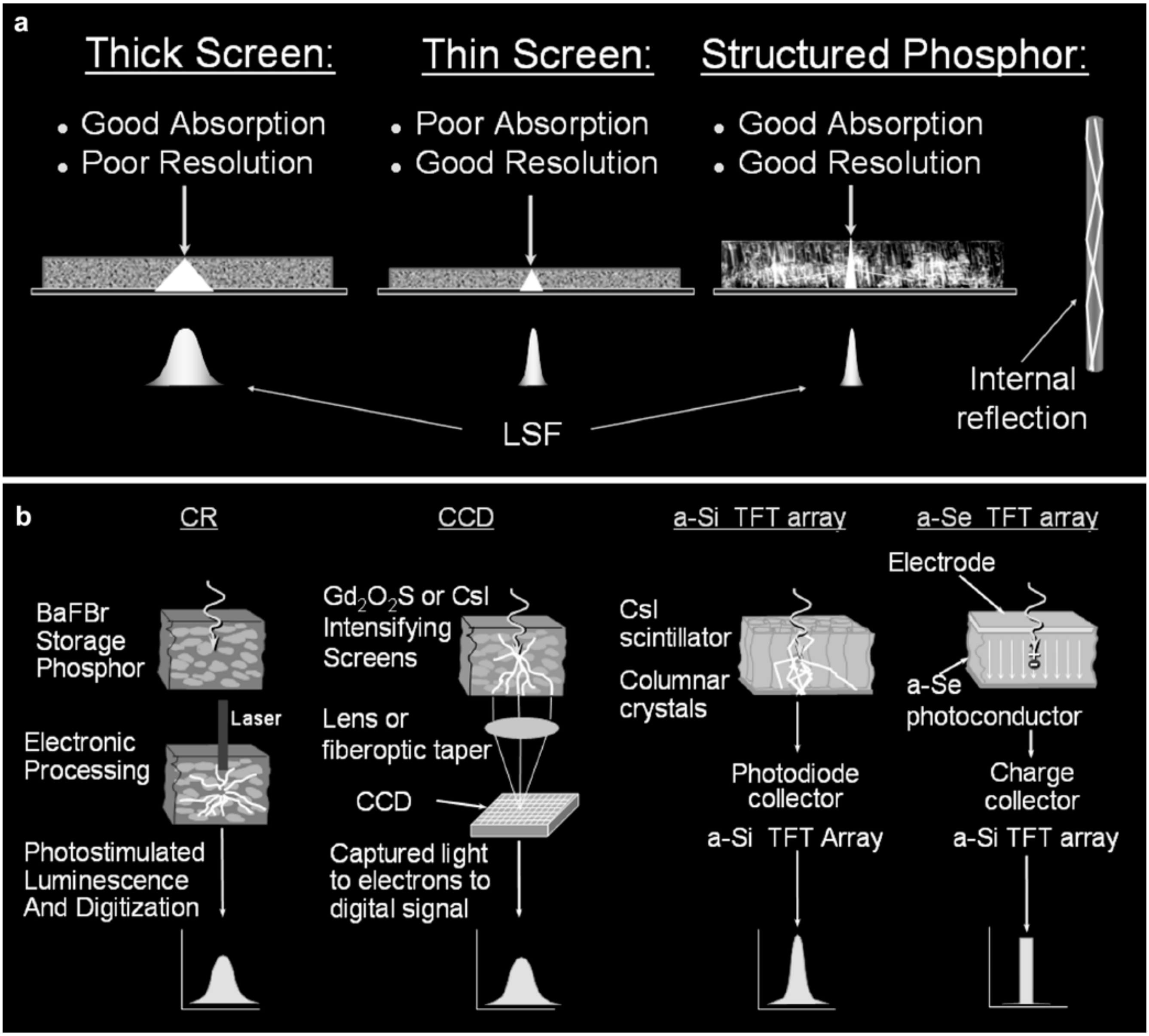

Fig. 2 Unsharpness as a limitation on detector efficiency varies with x-ray-to-light conversion technology. **a** With unstructured phosphors, blur is strongly dependent on phosphor thickness. Very thick structured phosphors manifest less blur because of internal reflections that channel light to the digital readout array. **b** Acquisition strategies for x-ray digital detectors are illustrated for CR, optically coupled CCD, amorphous silicon TFT array coupled to a structured CsI scintillator, and a TFT array coupled to an amorphous selenium semiconductor with direct x-ray to charge conversion. Line-spread functions are approximate and do not indicate actual performance

size and spatial frequency are inversely related, meaning that high spatial frequency represents small objects and minute detail, while low spatial frequency represents large, coarse objects. A perfect system has an MTF of unity at all spatial frequencies. A typical digital imaging system has an MTF that decreases with increasing spatial frequency. The effective detector element size (often termed the pixel, but more appropriately described as the detector element or del) is the corresponding area on the detector over which smaller signals are averaged, and represents the maximum resolvable detail that can be transferred by the system. This is known as the cutoff frequency, which is equal to the inverse of the dimension, typically in inverse millimeters; for example, a detector element size of 0.1 mm (100 μm) has a cutoff frequency of $(0.1\ \text{mm})^{-1} = 10\ \text{mm}^{-1}$ (this is ten cycles per millimeter). Another important consideration for discretely sampled image detectors is the concept of signal aliasing and the Nyquist frequency, which states that the maximum useful signal requires two

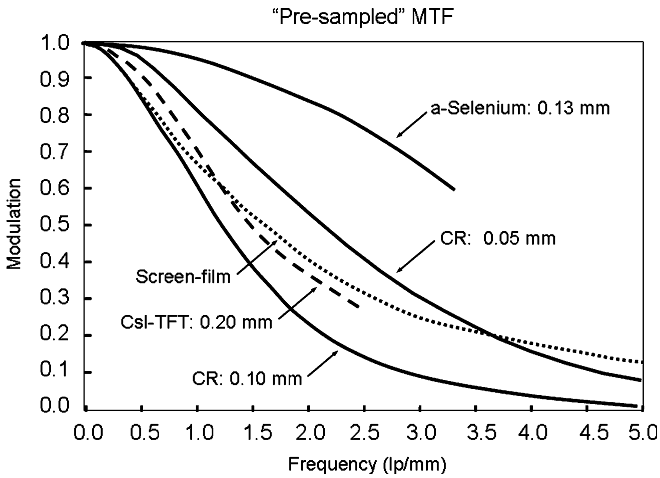

Fig. 3 The pre-sampled MTF curves (acquired in a manner to eliminate the effects of the detector element size) are illustrated for various digital detectors, and compared to a 400 speed screen-film (*short-dash line*). For each detector, the element detector is also indicated. The *curves* end at the Nyquist frequency value determined by the detector element size

samples per cycle, and the Nyquist frequency = $(2 \times \text{dimension})^{-1}$. For the same 0.1 mm dimension, the Nyquist frequency = $(2 \times 0.1\ \text{mm})^{-1} = 5\ \text{mm}^{-1}$, or ½ cutoff frequency. When object frequencies are contained in the object spectrum above the Nyquist frequency, this signal is reflected back into the lower spatial frequencies as aliased signals and unwanted noise.

Different detectors have different modulation transfer characteristics, as shown by Fig. 3. The chief determinant of resolution is the information spread that occurs as the absorbed x-rays are converted into secondary light photons. While some detector systems exhibit superior MTF response and have superior rendered detail, the overall image quality might be less, as information-transfer efficiency is also important.

X-ray quantum mottle

The SNR is a measure of the average value of a signal relative to its background noise. A more pertinent measure for a detection task is the CNR or detail SNR, which is a measure of the contrast derived from the image based upon attenuation difference (the subject contrast) relative to the background noise. Contrast is described as the fraction of a signal relative to the brightness, B, of the background, as $C = \frac{\Delta B}{B}$, and is also a function of the area (diameter) of the signal. There are many sources and types of noise that must be considered. First and foremost is the noise associated with the image formation process – that is, the detection and conversion of primary x-ray photons that are transmitted through the object into a visible image. The signal, S, is proportional to the number of photons absorbed in the detector: $S = N_0\, \eta g$, where N_0 is the incident number of primary photons, η is the efficiency of detection, and g is the gain of the detector. It can be shown that for x-ray quantum limited detection, the number, N_0, of x-ray photons per unit area absorbed and converted by the detector into a visible image are Poisson-distributed, with the variance, $\sigma_q^2 = N_0\eta(g^2 + \sigma_g^2)$, where σ_q^2 is the variance of the image, and σ_g^2 is the variance of the detector gain. The ratio of the signal and noise is calculated as $SNR^2 = N_0\eta \frac{1}{1+\sigma_g^2/g^2}$; Thus, the SNR is proportional to $\sqrt{N_0\eta}$, and the SNR increases as the number of x-ray photons incident on the detector increases as long as the gain noise terms are not too large.

When a contrast target of area A in a quantum-limited noisy background is tested for detection by a human observer, a threshold contrast-to-noise ratio, k, of approximately 5 is necessary to reliably detect the object based upon human visual perception studies [3]. From the viewers' perspective, the visibility of the signal is dependent not only on the contrast but also on the target area. For circular targets (commonly used in threshold detection experiments), the area is proportional to the square of the diameter, $A \propto d^2$. In terms of reliable detection of a contrast target for an ideal photon noise-limited detector it is thus $N_0\eta\, d^2C^2 = k^2$. The left side of the equation is the number of photons producing a signal per unit area of the contrast target, equated to the ability to reliably detect the signal within the given area. Taking the square root of the expression yields: $k = \text{SNR} \times d \times C$, which predicts that the reliable detection of the test objects diminishes linearly with a linear decrease in SNR (square root of the incident photons or radiation dose), target diameter, or target contrast. CNR has meaning only when the size of a test element is specified, and the resolution of a system has meaning only when the contrast of the test element is specified. Contrast-detail phantom detection tasks (see the image illustrated in Fig. 4) are based upon this relationship, which demonstrate the capability of a specific radiation detector to qualitatively and quantitatively detect a contrast signal of a given contrast and size in a noisy background for a known incident exposure to the detector. The minimal detectable contrast is often insufficient to make a reliable diagnosis; in fact, a significantly higher SNR, minimally on the order of 10 to 15 (personal observation from radiologist feedback) is necessary to provide the minimum acceptable image quality and information content as deemed necessary by radiologists.

Other noise sources (Table 1)

X-ray imaging detectors are hardly ideal, however. X-ray scatter from the object (patient) will diminish the

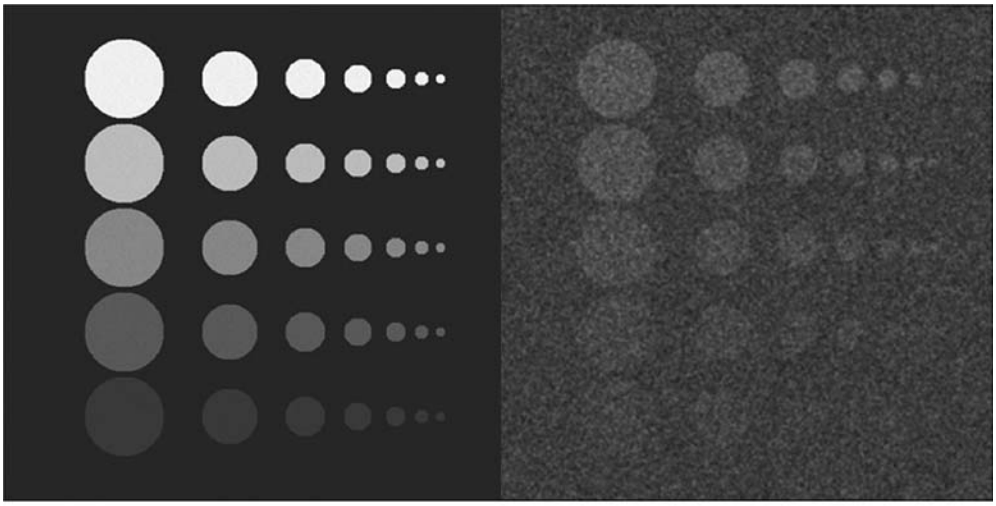

Fig. 4 *Left*: Contrast detail phantom of diminishing diameter from left to right, and diminishing attenuation *from top to bottom*. *Right*: Image of C-D phantom for a specific incident exposure. Visibility is shown to be a function of object area and object contrast relative to the background noise, as explained in the text

Table 1 Secondary noise sources that reduce CNR

Scatter
Detector thickness variations and structured noise
X-ray to light conversion noise
Swank noise
Spatial sampling and quantization errors
Aliasing
Antiscatter grid lines and out off
Errors in flat-fielding correction
Anatomical structure noise

subject contrast by adding signal to both the target and the background. If not appropriately filtered, the low-energy part of the polychromatic x-ray spectrum will deposit dose to the patient without transmitting a significant portion of primary radiation to the detector, reducing dose efficiency. Thickness variations and structured noise (e.g., scratches, dust, imperfections) in the detector substrate can lead to unwanted additive noise in the acquired image. Cascading processes such as x-ray to light conversion (gain noise) can introduce a significant variation exceeding that of the x-ray fluence, which will dominate the noise statistics. More efficient (higher gain) production of secondary carriers improves (decreases) the conversion noise. Some optically coupled digital detectors have a secondary quantum sink which results from the inefficient collection of light by the lens, leading to statistics that fall below that of the statistics of the detected x-ray photon fluence. Swank noise [4] results from the variation in light output as a function of absorption depth in a scintillator, adding to the overall noise. Spatial sampling and quantization errors occur during the conversion of a continuous analogue signal into a discrete digital signal. Aliasing is a phenomenon contributing to the detected noise in discretely sampled digital systems, and is most problematic in direct conversion detectors that have high signal-transfer modulation at the Nyquist frequency. If an antiscatter grid is used, grid cutoff or grid lines can inject unwanted noise into the image. A process called flat-fielding reduces noise sources caused by stationary detector non-uniformities with the acquisition of a uniformly exposed, low-noise image (or series of images) to identify and subsequently correct variations by applying a normalized and inverted correction image. (See the section on image pre-processing in the next section of this article.) Flat-fielding can be very successful, but requires frequent recalibration and sometimes is less successful when the image to be corrected does not match the exposures of the flat-field image. Another source of noise, also known as anatomical structure noise, is caused by the superimposition of unwanted anatomy over the area of interest, intrinsic to the 3-D patient volume projected onto the two-dimensional image plane. This source of noise is largely uncorrectable except in some cases where temporal subtraction or dual energy subtraction can be employed (albeit at a higher dose), or in careful patient positioning that can cast unwanted anatomy away from the area(s) of interest.

All additive noise sources can reduce the overall contrast-to-noise ratio in the image, and thus reduce image quality, in spite of high incident x-ray exposure levels. This means that many noise sources cannot be overcome by increased exposure to the patient, and must be controlled by other means. Achieving ALARA radiation doses in pediatric radiology requires acknowledgement and understanding of all noise sources (not just x-ray noise), and methods for reduction or elimination by the proper selection and/or use of digital x-ray equipment.

Objective measures of x-ray detector performance

Imaging physicists use objective methods to determine detector and x-ray system performance (image quality) and efficiency (dose). Measurements include signal-transfer capabilities (the modulation transfer function described previously), noise-transfer characteristics (the noise power spectrum, NPS), equivalent noise metrics (noise equivalent quanta, NEQ), detective quantum efficiency (DQE), which is a measure of the signal-transfer efficiency of the detector system, and receiver operator characteristic (ROC) curves [5]. Except for the latter, these measurements are a function of spatial frequency, meaning that the signal- and noise-transfer characteristics are dependent on effective object size. An ideal imaging system would have 100% MTF at all spatial frequencies, and NPS / NEQ indices determined strictly by x-ray quantum statistics incident and totally absorbed and converted to a useful signal by the detector, resulting in a 100% DQE at all spatial frequencies.

The response (e.g., the digital number output) of a digital detector to incident exposure is the characteristic curve. Most digital detectors have a linear response to variations in incident exposure, and those that do not (e.g., logarithmic response of CR systems) can be linearized with respect to incident exposure. Under linear conditions, the MTF and NPS values can be measured and the detective quantum efficiency (DQE) can be calculated as a measure of the signal transfer efficiency for a given incident exposure as a function of spatial frequency, defined as:

$$DQE(f) = \frac{SNR^2_{\text{out}}}{SNR^2_{\text{in}}} = \frac{< PV >^2 \times MTF(f)^2}{NPS(f) \times \phi} = \frac{NEQ(f)}{\phi}$$

In the above equation, the $\mathrm{SNR}^2{}_{\text{in}}$ is equal to the incident fluence (number of x-ray photons per unit area on the detector), and the $\mathrm{SNR}^2{}_{\text{out}}$ is the square of the measured SNR of the output signal over a large uniform area. The DQE(f) is quantitatively determined as a function of spatial frequency by the measurement of the average global pixel value $<\mathrm{PV}>$, the MTF(f) representing the signal transfer characteristics, and the NPS(f) representing the noise transfer characteristics as defined by specific methodologies [5–7]. NEQ(f) is a measure of the effective signal to noise characteristics of the detector as a function of spatial frequency. The DQE(f) measurement is typically performed over a range of incident exposures commonly used for clinical applications to determine incident exposure dependencies of the detector. Ideally, the DQE (f) is 100% at all useful spatial frequencies for all incident exposures, but in reality is typically less than 80% at DQE(0) for the best digital detectors, and drops rapidly with increasing spatial frequency, as all detectors suffer from many, if not all of the additive noise sources and loss of spatial resolution. DQE(*f*) is also dependent on the Quantum Detection Efficiency (QDE) of the detector (these two characteristics are often confused or considered the same, but this is not a typographical error). The QDE is the fraction of incident x-ray photons absorbed in the detector, and therefore sets the upper limit of x-ray utilization efficiency, and is not a function of spatial frequency.

Comparison of digital detectors based upon objective metrics such as MTF, NPS, NEQ, and DQE allows one to determine the efficiency of signal transfer as a function of spatial frequency (Fig. 5). In addition, the DQE can be used to estimate the relative incident dose on a detector needed to achieve a given SNR in the output image. For example, at 0.5 lp/mm (a relatively low frequency), the dose to a conventional CR detector compared to a CsI-TFT array detector to achieve the same SNR would be about 0.65/0.25 = ~2.6 × more, to compensate for lower efficiency of information transfer.

Image pre- and postprocessing

Since the DQE measures how efficiently the input X-rays are used in the presence of a uniform field and other stochastic (e.g., electronic, digital) noise sources, then a given image SNR (loosely indicating image quality) is more efficiently achieved with a system having a higher DQE. A digital radiography system with higher DQE, however, does not necessarily translate into a system with superior image quality. The system is composed of a number of subcomponents, one of which is the x-ray detector, but others are the x-ray system, x-ray energy and beam uniformity, wide-latitude response of the digital detector,

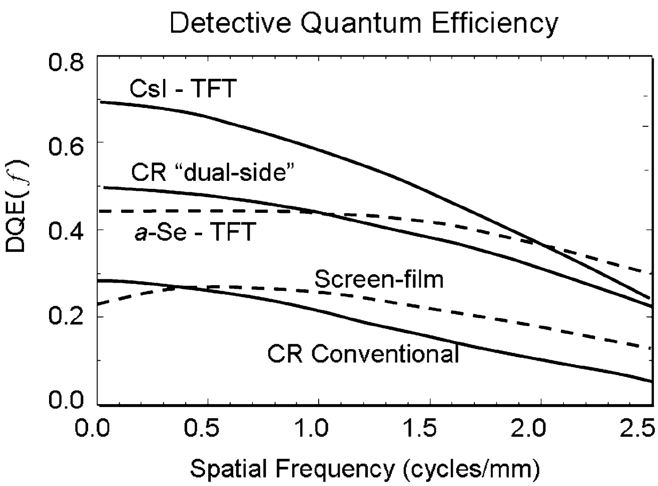

Fig. 5 Approximate DQE for various detectors is extracted from a review of the imaging physics literature out to a spatial frequency of 2.5 mm^{-1}. Note: this is a generalized indication of technology capabilities in 2004

optimal spatial and temporal sampling, proper image processing for the specific anatomy or disease process of interest, image display quality and calibration, viewing conditions, and radiologist alertness and working conditions. From the perspective of the output image, the image pre- and postprocessing, monitor grayscale calibration, and viewing conditions are critical applications that could mean the difference between a diagnostic and nondiagnostic image.

Raw data collected by a digital detector are fraught with detector non-uniformities, dead pixels, amplification differences of submodules, light collection efficiency of secondary acquisition devices, and geometric distortion, among others. For fixed, stationary variations, corrections are implemented in a preprocessing stage using either one-dimensional (for line-scan detector systems such as CR) or two-dimensional (for CCD or TFT-array detectors) correction techniques commonly known as flat-fielding [8]. The idea is to expose a detector to a noiseless (high exposure averaged over many image acquisitions) and uniform x-ray fluence. Collected image information is measured, normalized, and inverted to form a flat-field image, which is subsequently applied to uncorrected clinical images in order to remove the inherent distortions, variable gain, and static noise, as shown in Fig. 6. This procedure effectively reduces or eliminates many sources of geometric inaccuracies and stationary noise that would otherwise be amplified by the gain and postprocessing stages of image output conversion. Adequate flat-fielding requires

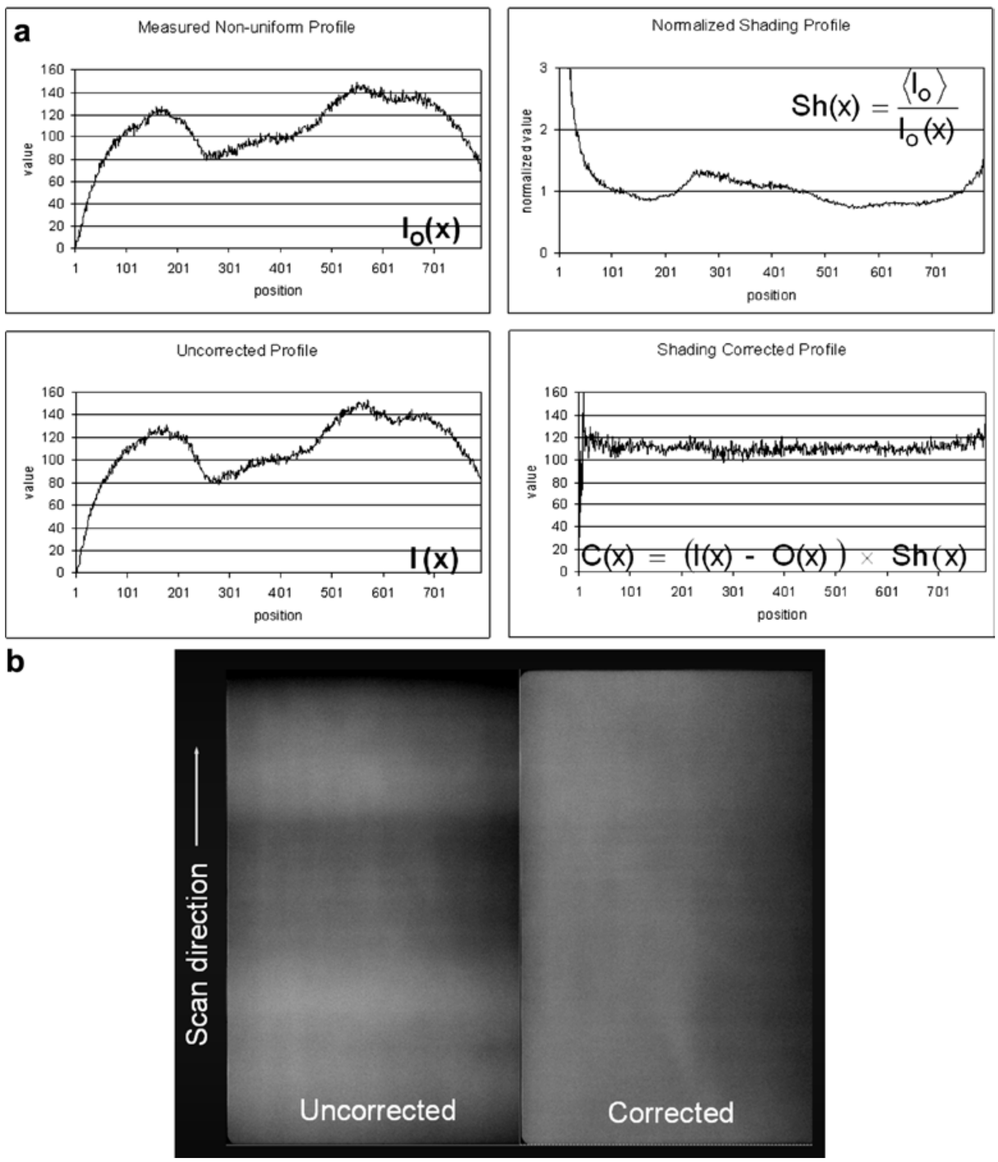

Fig. 6 Flat-field technique. **a** Illustration of a one-dimensional flat-field technique to reduce stationary noise sources in a digital CR detector. **b** The corresponding uncorrected and corrected images are shown. Note that the shading correction is applied in the scan direction, indicated by the *vertical arrow*. (From [8] with permission)

knowledge of the methodology and frequency of calibration as recommended by the manufacturer, as digital detectors have unique methods and requirements for flat-fielding.

Optimized image post-processing is also crucial to achieving the optimal pediatric radiograph.

Optimized image postprocessing is also crucial to achieving the optimal pediatric radiograph. Raw image data must go through several stages of processing, chiefly due to the wide-latitude response of the digital detector and the information contained in the latent image. The first stage is exposure recognition and segmentation of the pertinent image data. These steps find the image data using histogram analysis to identify the minimum and maximum useful values. The second stage applies scaling operations to ensure that the range of valid histogram values is shifted according to the exposure to the receptor (i.e., if the receptor is underexposed, the raw data will be shifted by amplification to an appropriate data range). Noise-reduction algorithms are often applied at this point on the raw data, adapting averaging schemes to reduce noise according to anatomy based upon image intensity. The third stage is the transformation of the scaled, raw digital values through a nonlinear relationship (e.g., a characteristic curve that mimics film response or predefined anatomy-specific processing) for contrast enhancement. By performing the first and second steps correctly, the useful range of the raw, scaled data is consistently mapped to the appropriate brightness and contrast renditions for grayscale enhancement. Often, this auto-scaling leads to the misconception that any digital detector can provide a lower dose image than a corresponding screen-film image. However, for many digital detector systems with lower detection efficiency, this is not true, because the system amplifies not only the reduced signal, but also the corresponding quantum and detector noise, with the net result a correspondingly low SNR due to insufficient numbers of absorbed x-ray photons. On the other hand, and potentially more problematic, is the overexposure condition, where the image looks very nice, but the patient is needlessly subjected to too much radiation.

The fourth step is edge restoration and enhancement to compensate for image blur caused by the x-ray tube focal spot with object magnification, and the intrinsic resolution limits of the digital detector caused by signal spread in the scintillator or averaged over the detector element area. With knowledge of the MTF response of the detector, this can be done in such a way as to limit the enhancement of noise relative to the signal. As the signal decreases with higher spatial frequency, but the noise remains relatively constant, edge enhancement processing must be performed carefully, and constrained to the low and intermediate frequencies. Often, a small spatial frequency range is chosen using a process known as blurred mask-subtraction. This process uses a blurred copy of the image, obtained by averaging adjacent pixels over an area with known weighting called a kernel, and subtracting from the original image [9]. The difference image represents a frequency bandpass which can be amplified and added back to the original image, providing an edge enhancement. Recently introduced algorithms such as MUSICA [10], MFP [11] and EVP [12] construct several different ranges of frequency bandpass information, each of which can be selectively amplified or de-amplified, and summed to obtain *simultaneous* image contrast equalization and edge enhancement. This image processing optimization has proven critical in the acceptance of CR at a large children's hospital [13] as shown in Figure 7, where standard processing was found to be inadequate in many pediatric examinations.

Overall, these image post-processing procedures require specification of several control parameters, based upon noise control, latitude adjustment, edge enhancement, contrast enhancement, and grayscale rendition. At the outset, these parameters require careful tuning on an exam-by-exam basis so that radiologist preference in depicting grayscale and spatial frequency enhancement is consistently achieved. As systems become more sophisticated and users more experienced, the expectation is that there will be a number of disease-specific image processing algorithms available at the push of a function button to allow the radiologist several different renderings of the displayed anatomy to increase the conspicuity of a possible lesion or anatomical characteristic, with fast sequential or simultaneous viewing of the original image for comparison.

A significant and often weak link of the imaging system is the display component, where the image information is transferred to the human eye/brain system. Soft-copy display calibration to the perceptually linear response curve is recommended per the DICOM Part 14, Grayscale Display Function standard [14]. Display optimization involves a specific calibration procedure to determine the response of the monitor to a range of digital driving levels (digital number in, display luminance out) with the subsequent transformation of optimized digital numbers to drive the monitor output according to the perceptually linear curve. This, in addition to proper illuminance (background lighting) conditions for the reading area is necessary, as the human visual system interface is also a key aspect of the total image quality optimization process as well. The American Association of Physicists in Medicine has a document in the final stages of acceptance for publication that details the display calibration and quality control methodology for soft-copy image display monitors [15].

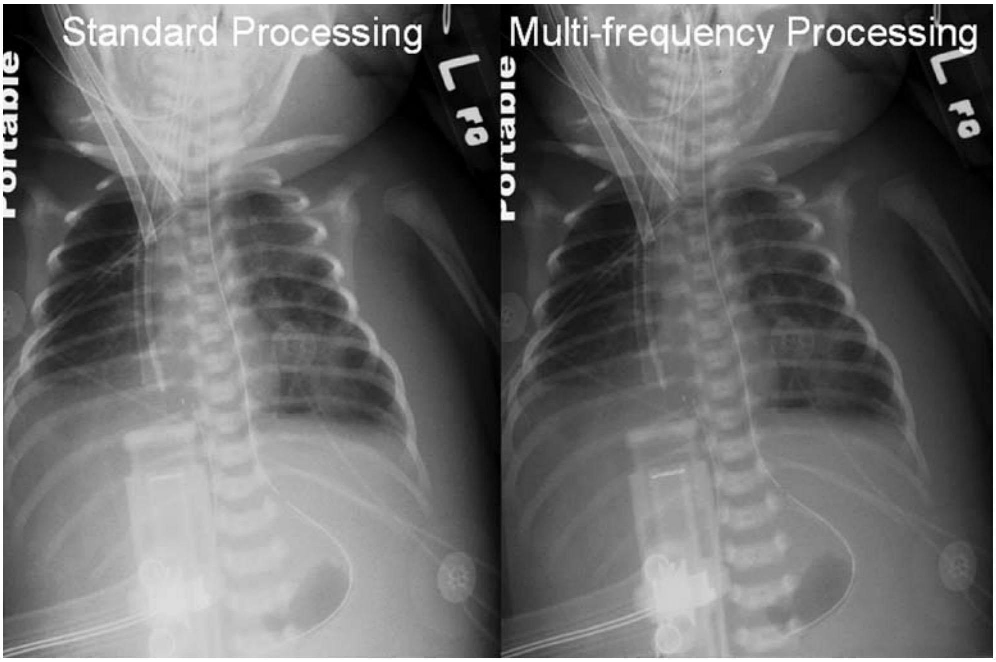

Fig. 7 Comparison of a suboptimally processed CR image using conventional methods (*left*) and the same image with multi-frequency processing methods (*right*) shows the importance of optimized image processing. Note better visualization of the airway, retrocardiac, and subdiaphragmatic regions. (Images courtesy of Keith Strauss, MS, Boston Children's Hospital)

Image quality and radiation dose

So, the question to be answered is: how does one achieve the appropriate image quality, fidelity, and intelligibility at the lowest possible dose? To answer this requires an investigation of the various digital radiographic detectors that are used for pediatric imaging, x-ray acquisition techniques employed as a function of patient thickness and mass, the image-processing algorithms applied to the images to enhance detail and display contrast, and the manner in which the images are displayed to convey the maximal information to the human viewer. Another issue is that of training. Radiologists are trained to find specific stimuli in noisy backgrounds; their experience overcomes some of the interfering noise patterns that would otherwise make a stimulus invisible, which could potentially result in a missed diagnosis for one observer, while another observer would detect it; thus another variable is that of the observer.

Regarding digital radiographic detectors, those systems with a higher DQE can produce images exhibiting a necessary SNR at a lower dose. Comparison of the ratio of DQE values at a specific spatial frequency will indicate the approximate dose savings possible. For example, in Fig. 5 the DQE curves show a decided advantage to the CsI-TFT detector at low spatial frequencies, which can result in 1.5 to 3 times lower dose compared to other digital detectors in order to achieve the same image quality. At higher spatial frequencies, however, cross-over of the DQE curves show that the a-Se direct digital detector has a higher DQE, and can be more radiation-dose efficient for high-resolution exams (e.g., pediatric bone fractures, abuse cases). Additionally, there are other considerations where fractional dose savings are of secondary importance, such as the need for flexibility in radiographic positioning (advantage goes to CR) and the acquisition cost of the systems (advantage to CR and some CCD-based digital detectors). With respect to technological capabilities currently available, all digital detectors can provide adequate image quality at a reasonably low dose when the overall system is optimized.

Computer simulations using Monte Carlo algorithms are often used to determine the optimal parameters for developing tissue contrast in terms of radiographic technique, patient characteristics, and radiation dose for a given CNR or detail SNR. Pediatric images are inherently of relatively low contrast; thus, lower kVp settings are often used with screen-film detectors in order to achieve the optimal contrast-to-noise ratio. Digital detectors are noise-limited in operation, meaning that postprocessing can provide display contrast that is chiefly limited by the magnitude of the noise. Optimization of radiographic techniques take into account the absorption of x-rays in the body (dose) as a function of the input spectrum (kVp and filtration) for the number of transmitted x-ray photons absorbed in the detector and required to achieve a given CNR or detail SNR. (Note that the absorption characteristics of the detector must also be considered.) Researchers often optimize a value calculated as the square of the contrast to noise ratio relative to absorbed dose: CNR^2/dose, which is a unitless value called the figure of merit (FOM), as one measure of determining the optimal kVp with low dose as the constraint and desired goal. This value is unitless because the CNR is proportional to the square root of the incident exposure (or dose), and therefore the CNR^2 is proportional to dose, and the ratio cancels units.

In general, what is found for pediatric patients is that the optimal kVp is chiefly dependent on the thickness of the patient and the tissue being considered (e.g., bone vs. soft tissue). The optimal kVp is usually higher for bone-to-soft tissue contrast and for thicker objects. Although low kVp increases the tissue contrast in the transmitted beam, there is a practical lower limit determined by the number of information-carrying x-ray photons (primary x-rays) that are transmitted without interacting in the body versus those that are absorbed or scattered. Longer exposure time needed for low kVp techniques can result in unacceptable image quality due to patient motion. Sometimes with neonates and small children, low kVp is used so that the minimum exposure time (which can be rather high for a portable x-ray unit) does not result in overexposure. This is where added tube filtration should be considered (e.g., 0.1 mm Cu). At the other extreme, high kVp reduces the inherent subject contrast of the tissues in the transmitted x-ray beam, but since a greater fraction of x-rays passes through the body without interacting, a much lower overall absorbed dose is achieved with a shorter exposure time. Even though contrast is compromised, with digital image processing much of this loss can be compensated by window and level adjustments to enhance the display contrast. This is acceptable as long as the effective beam energy is not too high within the constraints of the image SNR. Certainly, there are practical limits that must be considered on the high end as well. Dedicated imaging equipment tuned for specific patient examinations is the ideal situation, but often not possible due to economic constraints. For larger patients, the tube filtration might be excessive relative to the output capabilities of the x-ray system, and must be removed, which confronts the technologist with many decisions in a situation that demands utmost attention to dealing with the patient. The bottom line is that there are a number of factors that must be considered on an exam-by-exam basis to determine the optimal kVp, mA and exposure time (mAs), and tube filtration, based upon patient thickness, mass, and the required contrast-to-noise ratio for a given detector DQE. x-ray technique is an area that has received much attention for pediatric CT examinations [1], and is certainly important for the optimization of radiographic exams. Technical factors are dependent on the exam type, patient size, detector absorption characteristics, and optimal contrast-to-noise ratio (lowest dose) compatible with performing the examination.

The challenges of pediatric imaging are immense, mainly due to the extremely large differences in patient size in terms of thickness and mass. This requires precisely calibrated radiographic equipment and the establishment of radiographic techniques based on physical size for a particular examination. Techniques can span a range of 100 or more mA for a 70 kVp beam in comparing what is required to image a neonate without a grid versus a large child using a grid. Tube filtration, as discussed previously, is also an issue. While not recommended for smaller patients, a scatter reduction grid should be used for larger patients and its attenuation appropriately considered in the development of the technique charts. Use of gonadal shields is always recommended when possible, but their use can alter the appropriate scaling of image data because the image numbers (histogram distribution) are altered by the presence of the shield. Uncooperative and distraught children dictate use of a short exposure time. Pediatric disease states can mimic radiographic noise, so optimal and consistent image acquisition and processing is essential in order to distinguish physical disease from poor image quality. Because digital detectors produce less limiting spatial resolution compared to screen-film detectors, particular care must be exercised when evaluating minute fractures and potential child abuse cases. These latter situations might require a higher dose to achieve the necessary SNR to make the diagnosis. A generalized systems consideration for image acquisition, display and interpretation illustrated in Fig. 8, and image quality/dose tradeoffs are listed in Table 2.

ALARA examples of appropriate image quality and dose

Practical methods of achieving ALARA are obtained through careful clinical practice. Radiographic techniques for all pediatric examinations should be posted and consistently employed in the diagnostic imaging areas. This requires a significant effort in adjusting and verifying x-ray tube outputs and beam quality for all systems in a department, so that it doesn't matter which imaging system is used. Radiographic examinations do not necessarily require the same SNR/CNR as others. For instance, pediatric scoliosis exams can often be obtained at a much lower dose when there is a simple spine measurement requirement as opposed to a diagnostic evaluation (see Fig. 1). Therefore, it would be considered ALARA if a longitudinal scoliosis study were performed initially with a higher exposure and lower quantum mottle (e.g., a diagnostic incident exposure of 1.0 to 0.5 mR), equivalent to a screen-film detector speed of 200 to 400, and subsequent followup studies performed at a much lower incident exposure of 0.15 mR or lower (e.g., 1000 to 1200 equivalent speed), which might be adequate for spine measurements using the higher-contrast bone landmarks that are appropriately visible at three times less dose. Certainly, in this case one might want to periodically perform a

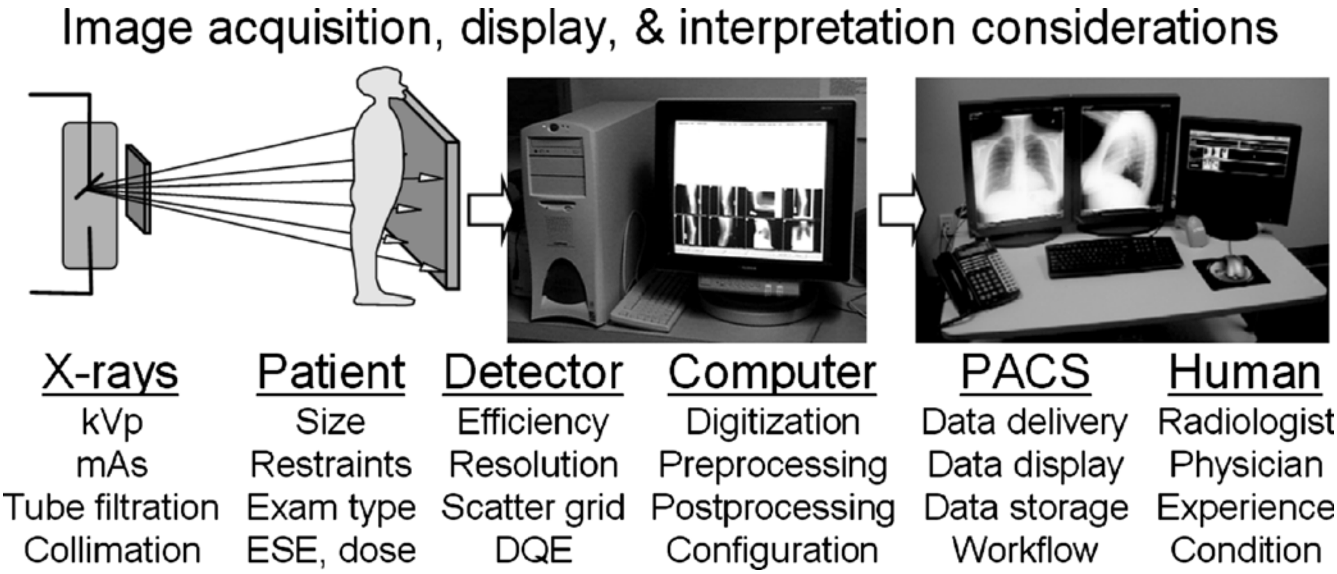

Fig. 8 Image quality and dose tradeoffs are a consequence of a number of considerations that require optimization from the acquisition, patient handling, detector, computer, image processing, display, and interpretation steps

lower-speed (higher dose) study, but a significant dose savings can potentially be obtained without any adverse affect on the diagnosis or utility of most images. Another example is a protocol to perform chest x-rays in the PA position when possible, such that radiosensitive tissues such as the breast receive a much lower dose than in a standard AP projection because the intervening tissues attenuate a significant fraction of x-rays in the intervening tissues. In a similar vein, it would be prudent from the ALARA perspective for longitudinal studies of the lateral chest x-ray, to acquire a left lateral projection for one examination, and a right lateral projection for the other, thus distributing the dose more evenly to the sensitive organs of the patient and keeping the peak dose as low as possible.

Table 2 Considerations for the tradeoff of image quality and dose

Issue	Characteristic	Consequence
Subject contrast	(increase image quality)	
	Low kVp	Higher dose
	High SNR	Higher dose
	Grid use and scatter rejection	Higher dose
Spatial resolution	(increase image detail)	
	Smaller pixel	Higher dose
	Low detector fill factor	Higher dose
Radiographic techinque	(Optimize x-ray parameters)	
	More tube filtration	Lower dose
	High kVp	Lower dose
	Strict collimation	Lower dose
	High DQE detector	Lower dose
Patient motion	(Reduce motion artifacts)	
	High kVp	Motion reduced
	Positioning aids	Motion reduced
Detector	(Considerations)	
	High DQE	CsI scintillator (structured)
	High resolution	a-Se photoconductor (direct)
	Flat-field calibration	High DQE, lower dose
	Cost	CR - $, DR - $$

Conclusions

Optimal image quality at the lowest possible patient radiation dose in pediatric radiography is achievable with a careful understanding of the factors that contribute to image quality and radiation dose, as well as thoughtful implementation of clinical routines, a-priori technique factors, examination-specific SNR requirements, and patient positioning. With the increasing use of digital radiography detectors, the disconnect between image appearance and under-/overexposures must be understood by not only the technologist, but the radiologist and physicist so that thoughtful and constructive feedback can lead to the optimal image quality at the lowest possible dose for such technologies. A variety of digital detector systems are available for implementation and application to pediatric radiology, all of which have advantages and disadvantages with respect to ease of use, absorption efficiency, spatial resolution, detective quantum efficiency, and cost, among many others. Because any imaging system is only as good as its weakest link, particular attention must be given to all of the cascading parameters that contribute to the final output image and its use, including exam-specific technique factors, patient positioning and geometry, proper use of an antiscatter grid (or non-use where appropriate), calibration and implementation of detector non-uniformity corrections, exam-specific image processing, correct image identification, adequate image display (on a calibrated image monitor), proper viewing conditions, and interpretation skill of the radiologist/referring physician. This long list is meant to point out that there are many instances during the image acquisition/display/interpretation process when image quality and associated dose can be compromised. Also important, therefore, is the continuous diligence to quality control and feedback mechanisms to verify that the goals of image quality, dose, and ALARA are always at the forefront.

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