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Management of pediatric radiation dose using Agfa computed radiography

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Abstract Radiation dose to patients and its management have become important considerations in modern radiographic imaging procedures, but they acquire particular significance in the imaging of children. Because of their longer life expectancy, children exposed to radiation are thought to have a significantly increased risk of radiation-related late sequelae compared to adults first exposed to radiation later in life. Therefore, current clinical thinking dictates that dose in pediatric radiography be minimized, while simultaneously ensuring sufficient diagnostic information in the image, and reducing the need for repeat exposures. Dose management obviously starts with characterization and control of the exposure technique. However, it extends farther through the imaging chain to the acquisition system, and even to the image processing techniques used to optimize acquired images for display. Further, other factors, such as quality control procedures and the ability to handle special pediatric procedures, like scoliosis exams, also come into play. The need for dose management in modern radiography systems has spawned a variety of different solutions, some of which are similar across different manufacturers, and some of which are unique. This paper covers the techniques used in Agfa Computed Radiography (CR) systems to manage dose in a pediatric environment.

Keywords Computed radiography · CR · Dose · Pediatrics · Imaging

Introduction

Although there is evidence in the literature of a health benefit from low levels of radiation exposure, an effect called radiation hormesis [1], other data suggest quite the opposite [2]. Current clinical practice is based on the latter body of data, and dictates that the radiation dose to patients be as low as reasonably achievable (ALARA), while still providing image quality adequate to enable an accurate diagnosis. This tenet imposes a high level of responsibility on the medical personnel involved in ordering and performing patient examinations, and on the manufacturers of the equipment that creates the images used for diagnosis.

What constitutes "adequate" image quality is open to discussion, but some level of consensus is emerging [3]. Interestingly, the absolute image quality of screen/film radiographic acquisition systems has actually decreased systematically over the past 30 years, largely due to the demand for more sensitive (i.e., higher-speed) detectors that allow patient dose reduction. This decrease, however, does not appear to have had any impact on radiologists' diagnostic accuracy or error rates. The advent of newer, more dose-efficient digital acquisition systems, like needle-phosphor-based CR [4, 5] and direct and indirect flat-panel digital radiography (DR) detectors [6] promises to deliver substantially higher image quality than current systems, while maintaining or even lowering patient dose.

While the acquisition system clearly plays a dominant role in determining the image quality achievable in the final displayed radiograph, image processing is also a critical component. Some modern image-processing techniques are capable of adjusting the level and type of processing within an image based on certain global and/or local image characteristics, for example, noise. Since the quantum noise inherent in the exposure itself is a major contributor to the final image noise, and since this noise becomes more visible as the exposure is reduced (i.e., the signal-to-noise ratio, or SNR, decreases with decreasing exposure), these image-processing techniques can be used to compensate, at least partially, for the poorer image quality that generally accompanies dose reduction.

I will address pediatric dose management by exploring exposure technique, the image acquisition system, and image processing. In addition, other aspects of dose management, like system quality control procedures, dose information management, and special procedures will also be covered.

Dose management during exposure

Patient exposure is the event in the imaging chain during which the most significant impact on patient dose is possible, regardless of the acquisition system. However, with few exceptions, CR systems do not communicate directly with the exposure equipment (user console, generator, automatic exposure control (AEC)). Therefore, information about the exposure parameters for any image (kV_p, mA, s, filtration, source-image distance (SID), grid, etc.) must generally be entered into the CR acquisition system manually by the radiographer. This is an infrequent event in normal clinical practice.

A complicating issue is that there is a much broader spectrum of body habitus in pediatric imaging than is found in adults (from sub-kg infants to 70+-kg adolescents). As a result, the definition of "standard" exposure techniques is difficult, since these are highly dependent on patient age and weight [3]. While Agfa and other manufacturers can and often do offer guidelines on appropriate CR exposure techniques for specific (age/weight) patient groups, de facto they have little control over the actual technique used for a particular patient. Given the large pediatric patient variability, this is probably good.

However, there are some boundary conditions to keep in mind. The sensitivity or effective speed of the storage phosphor screens currently used in all CR systems varies with kV. Although the variation is smaller than that found in screen/film (S/F) systems, Agfa recommends keeping the kV in the range from about 55–115 kV to bound this speed variation. Agfa also recommends performing periodic sensitivity calibrations of the system with a CR quality control (QC) tool (see below) in order to maintain consistent system performance over time.

Increasingly, facilities are raising kV to lower pediatric dose [3, 7]. For acquisition systems with a narrow exposure dynamic range, like S/F, mAs must be adjusted precisely to maintain the proper exposure to the detector (this usually means adjusting mA, since the exposure time is already short to stop motion in often uncooperative, moving patients). For wide-latitude detectors, like CR, such exposure changes are easier to make. By using additional filtration to harden the beam (for example, thin layers of copper [3]), the entrance exposure in pediatric imaging with CR can be lowered by up to 40% for certain body parts [7] (John Lobick, Agfa Corporation, personal communication, to be published). Guidelines for other S/F exposure parameters, like reference dose values, focal spot size, generator and tube characteristics, grids, SIDs, and AEC settings, have also been published [3], most of which translate easily into CR.

The use of AEC is sometimes problematic in pediatric imaging. The ionization chambers may be larger than the (often moving) patients, they have a kV-dependent response, and they are sometimes integrated behind nonremovable grids. However, since the dose efficiencies of Agfa's CR systems and conventional S/F systems are comparable and the cassette structure similar, Agfa's initial recommendation for those who intend to use AEC for pediatric imaging is to start with the same settings as those used for S/F. With the support of Agfa's applications team, these can be tuned at any time according to local operational preferences.

Dose management in the image acquisition system

Any radiographic image acquisition system must interact with the aerial x-ray image (i.e., the image in space) emerging from the patient, and convert it into either a visible or a digital image [4, 8, 9]. For CR, this requires efficient absorption of incident x-rays by a storage phosphor screen, or image plate (IP), and their conversion into a stable, exposure-linear latent image signal. This latent image signal must be easily (laser-) stimulated, and must release sufficient light to be measurable with a photodetector. Agfa's CR systems fulfill all of these criteria.

One unique feature of Agfa CR systems is that, unlike other commercial CR systems, which quantize (i.e., digitize) the analogue latent image signal coming from the photodetector in logarithmically spaced increments, Agfa uses square-root quantization [10]. Since the noise in quantum-limited (i.e., dose-limited) systems is proportional to the square root of the exposure (so-called Poisson statistics), this technique effectively makes the noise in the digitized signal independent of the signal

value (so-called additive Gaussian noise). In other words, the noise contribution from the x-ray exposure becomes fixed, regardless of signal level. This has important implications for the image processing stage (see below), and also provides higher signal (contrast) resolution at low exposures.

Current flying-spot CR systems (i.e., systems in which a laser beam is scanned systematically across the IP surface to read the latent image) have raw imaging performance that is generally comparable to that of screen/film systems. This performance is typically measured in terms of an input/output relationship (sensitivity or contrast), sharpness (characterized by the modulation transfer function, or MTF), and noise (characterized by the noise power spectrum, or NPS) [11].

Increasingly, a measure that combines these individual metrics into one (spatial-frequency- and exposure-dependent) metric, called Detective Quantum Efficiency or DQE, has been used. DQE measures the ability of a system to maintain at the output of the acquisition system the signal-to-noise ratio (SNR) inherent in the radiation incident on the detector. An ideal system, one that does not degrade the incident SNR at all (that is, does not degrade the signal, does not add noise), has a DQE = 1.0. The DQE of any real imaging system is always less than 1. DQE generally decreases with increasing spatial frequency. In other words, it is harder to maintain the incident SNR at higher spatial frequencies. In effect, the system noise swallows the finer signal details. From a dose management point of view, DQE is important because systems with higher DQE can use less detector dose to achieve the same output image quality that a system with lower DQE achieves with correspondingly higher detector dose.

DQE plots for a current Agfa CR system using commercial (powder-based) IPs are shown in Fig. 1. DQE(0), the DQE value at zero frequency (i.e., for large uniform signal areas), is around 25-30% and decreases as spatial frequency increases. This is typical of modern CR systems. CR systems that use dual-sided reading, that is, those that detect emitted light from both sides of the screen, have shown somewhat better DQE results, at least at lower spatial frequencies [12]. Agfa is developing a new screen technology in which the conventional "particle in a binder" structure of today's powder screens is replaced by storage phosphors grown in needle form [4], similar to the structured scintillators used in image intensifiers and in some flat-panel DR detectors. The material characteristics of this new structured storage phosphor and the absence of a binder produce a significantly (2x) higher x-ray absorption. This, along with favorable optical properties (such as a sharpness improvement due to light-piping through the needles, and better layer uniformity), increases the DQE at low frequencies for this CR technology to about 0.6 (Fig. 1), comparable to the levels found in the latest DR flat-panel systems. This doubling of the DQE can be used to improve image quality at the same exposure used for current CR systems, or to decrease patient exposure by about a factor of 2 while maintaining current CR image quality, or some combination of these.

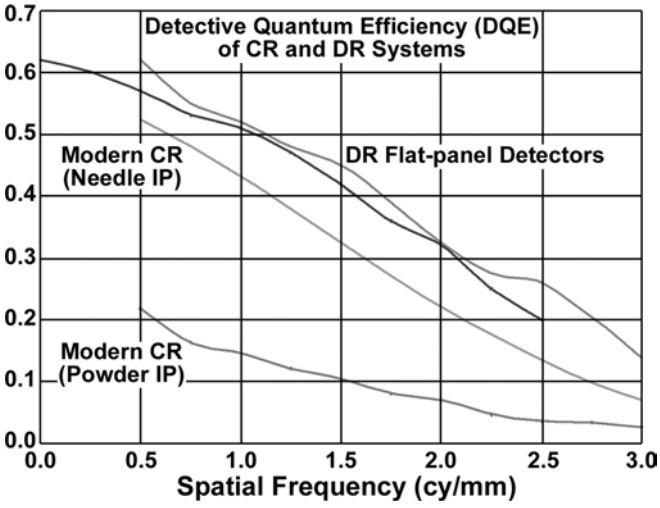

Fig. 1 The Detective Quantum Efficiency (DQE) of several modern CR and DR systems. The higher the DQE, the lower the dose requirement to achieve a given output image quality level. The (extrapolated) DQE(0) value for the current CR system (a flying-spot scanner using a powder IP) is around 25–30%. The next-generation CR system using a needle IP shows DQE(0) values about 2x higher than current CR systems. In fact, this needle-based CR system has image quality approaching that of modern flat-panel DR systems (DR data from Kump KS (2001) Fast imaging of a 41-cm amorphous silicon flat-panel detector for radiographic applications. Proc SPIE 4320:87–93, and Fasbender R, Schaetzing R (2003) Neue CR-Technologien für die digitale Radiographie. Radiologe 43:367–373)

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Regardless of the details of the acquisition system, a major issue is the estimation and control of dose. As noted, the CR acquisition system is seldom integrated with the exposing system, so direct communication of exposure technique and values is, at best, difficult. Thus, we are generally forced to extract or estimate exposure information from signals contained within the acquired image itself. This is an ill-defined problem in general (Fig. 2). Many combinations of patient body habitus and exposure can result in the same detected signal, so

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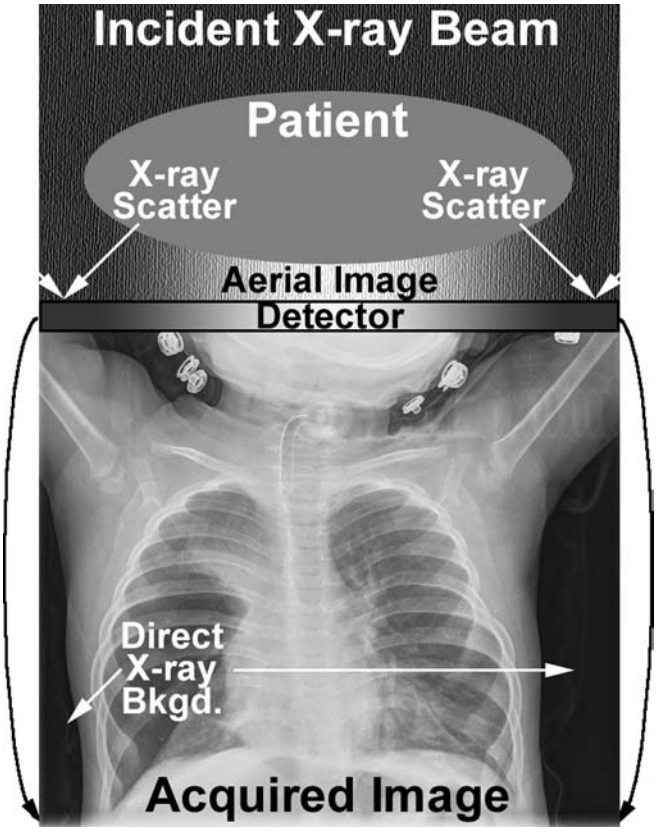

Fig. 2 It is generally impossible to estimate exposure and patient dose from only the detected image signal values, since many combinations of patient body habitus and entrance skin exposure (ESE) will lead to the same dose to the detector. Although the presence of direct x-ray background around the body part can help under certain circumstances, x-ray scatter from the body or other components in the exposing system can modify this direct signal in ways that are difficult to know or calculate. The average detected signal behind/under the body part provides a reasonably consistent dose estimate, and this is the metric currently used by CR manufacturers to provide dose feedback to the user

values extracted from the acquired image provide only a composite measure of patient and exposure.

A special case (Fig. 2) is when there is direct x-ray background imaged around the body part (not an infrequent occurrence in pediatrics). With knowledge of the input/output relationship of the CR system (i.e., detected signal vs. incident exposure), one might hope to extract from the signal in this area not covered by the patient an estimate of entrance skin exposure (ESE), from which one could infer some (averaged) information about patient dose. In this special case, this extraction has been shown to work reasonably well [13].

Unfortunately, not all images contain such a direct background region. In addition, x-ray scatter complicates the situation, since the background signal on the IP is an unknown combination of direct x-ray exposure and scatter from the body part or other parts of the system. Current CR materials are known to be more sensitive to

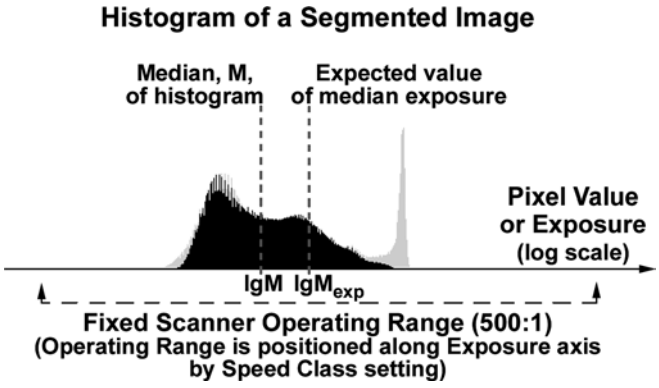

Fig. 3 Two histograms of the acquired image shown in Fig. 2. The *black histogram* is that of the segmented image, while the *gray histogram* is that of the raw (unsegmented) image. Analysis of the segmented histogram produces the dose feedback number, lgM, that indicates how close the average detector dose in some region of interest behind the patient was to the average detector dose expected from the speed class used for acquiring that image. In order to do this correctly, of course, the relationship between pixel value and exposure must be known (i.e., the system must be calibrated). See text for a description of lgM

x-ray scatter than S/F systems [14], a property attributed by some to the lower k-edge of the materials used in manufacturing most CR screens. The result: there is currently no robust algorithm to extract this kind of patient dose information accurately from the measured values on a CR plate. The problem is made even more difficult by the wide spectrum of pediatric body habitus mentioned earlier. We are therefore left with using the measured image values behind the body part as a surrogate for dose management. If we can maintain dose to the IP relatively constant at some value deemed to be appropriate for the exam or patient group of interest, then we have a way at least to track dose consistency in the CR environment. CR manufacturers provide different ways to do this.

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Agfa's dose feedback algorithm calculates a characteristic value of the histogram of the *correctly segmented* digital image (Fig. 3). The importance of proper segmentation cannot be overstated. Using the histogram of the raw, unsegmented image (i.e., the image including direct x-ray backgrounds, collimators) to estimate dose is fraught with danger. Images that contain large x-ray background areas will tend to produce artificially high average dose values, since these direct areas will expand the histogram to higher doses. Images with no direct background will produce a different average dose value,

even if the patient ESE values were exactly the same in these two cases. Dose values calculated only from data behind the body part will, in principle, provide for more consistent dose management. Thus, it is important to find (i.e., segment) on each image the location of the body part and create a dose estimate based only on the pixels within that area, ignoring any direct x-ray background or collimation.

For Agfa CR systems, the dose feedback is given as a so-called lgM value, which is the logarithm of the median value of the pixel histogram of the segmented image (Fig. 3). The median value was chosen because it is generally less sensitive to histogram outliers than are other statistics like mean and mode. A logarithmic rather than a linear metric was chosen to compress CR's large exposure dynamic range into a manageable number. The lgM value is always defined in the context of a Speed Class, a field in the operator interface. This concept familiar to users of S/F systems [15], except that CR's forgiving exposure latitude allows the nominal exposure for a given Speed Class to vary over a much wider range than in S/F (about 500:1 for each Speed Class). The Speed Class indicates to the system before scanning the approximate dose level expected at the detector, and enables the acquisition system to adjust itself for optimum detection. The lgM value calculated from the resulting image indicates how close the actual detector dose was to the expected dose and, by implication, how close the patient ESE was to the exposure that would have been used at that Speed Class in an S/F environment.

By definition, the target lgM value for any Speed Class is about 1.96. Because of its logarithmic nature, each change of roughly 0.3 in lgM corresponds to doubling or halving the dose. For example, if the lgM value for a given image is calculated to be 2.26, it indicates that the dose was about twice as high as that expected for the selected Speed Class. If the Speed Class had been 400, the expected detector dose, by definition [15], would have been about 2.5 μGy. A lgM value of 2.26 tells you that the detector actually received about 5 μGy instead of the expected 2.5 μGy.

Obviously, consistent, reliable dose feedback occurs only when the system has been calibrated properly (more on calibration and quality control below). In addition, lgM is sensitive to a number of other factors. The most critical one has already been mentioned: segmentation. Any errors in the segmentation algorithm (e.g., including background that does not belong to the body part, or excluding portions of the body part) can cause variations in lgM. In a related effect, collimation can also affect lgM. While tight collimation is a goal in pediatric imaging, the small body parts involved sometimes let direct x-ray background through, which, if not excluded by the segmentation algorithm, will tend to raise the lgM value. Another potential source of error is the delay between exposure and scanning. The latent image signal in storage phosphor screen decays very slowly over time, with the most rapid decay occurring within the first hour or so after exposure, so variations in scanning delay will result in different levels of signal remaining on the IP, even for the same ESE. Finally, the reproducibility of the exposing system will clearly be mirrored in the reproducibility of lgM.

The ability to get dose feedback from a CR system is clearly important, but one can build in additional functionality that focuses on the "management" part of dose management. Agfa's ADC dose monitoring software is such an example. This software enables each facility to set up reference or target dose values for up to 200 exam categories, either by decree or during a learning phase, in which the self-calibrating software registers exposure levels for 50 consecutive images of each category, and sets up an internal table of exposure norms. When the lgM of a newly acquired image deviates from the stored target or reference value for that exam category, the image is flagged. In addition, each output image contains a numerical and visual (bar graph) display of the lgM value relative to the reference value that shows the extent of over- or underexposure.

The software also maintains a history file containing dose (lgM) information for the last 50 exposures in each exam category so that radiologists or radiology administrators can monitor exposure consistency and investigate/correct any occasional or systematic deviations.

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Dose management with image processing

Managing dose with image processing appears improbable. After all, the exposure and acquisition have already occurred. However, modern image-processing techniques often have ways to adapt their processing to the local characteristics of the image, in particular to the noise content, which, as we know, is sensitive to dose. For example, by adjusting the contrast or the amount of enhancement based on the local noise level, it is often possible to achieve diagnostic image quality at lower dose levels. By taking these image-processing capabilities into account during acquisition, it may be possible to reduce dose.

An example of such software is Agfa's MUSICA (multi-scale image contrast amplification) [10, 16]. This technique decomposes the acquired image into a resolution pyramid with around a dozen layers, where each layer represents a sub-band of spatial frequencies or detail sizes present in the original. Based on extensive clinical studies, the algorithm processes groups of layers

in different ways according to a set of user-definable, exam-specific parameters. The unique needs of pediatric imaging have spawned a special add-on to MUSICA (ADC pediatric application software) that takes into account the sizes and contrasts of the relevant diagnostic structures in four different pediatric age/weight groups for a variety of exam types. A next-generation version of MUSICA will be able to adjust its processing to the image automatically, with no user intervention or input at all. This latest version is currently in clinical testing [17].

In its simplest form, MUSICA boosts the contrast of low-contrast objects in each sub-band, while reducing slightly the contrast of those details that are already quite visible. When the layers of the pyramid are recombined appropriately, low-contrast objects have improved conspicuity compared to the original, and high-contrast details are still easily visible. This process is guided by, among other things, the noise content of the image. Because of Agfa's use of square-root quantization in the acquisition process, the noise, which normally varies nonlinearly with signal, becomes a constant, the magnitude of which can be measured by the software in smooth regions of the image. In areas and frequency ranges in which the noise dominates the useful signal, contrast is not boosted. Areas/frequencies where the signal is sufficiently above the measured noise level are candidates for contrast amplification. Thus, the image processing can adapt its operation to the dose to produce optimized images. An example of MUSICA processing of a pediatric image is shown in Fig. 4.

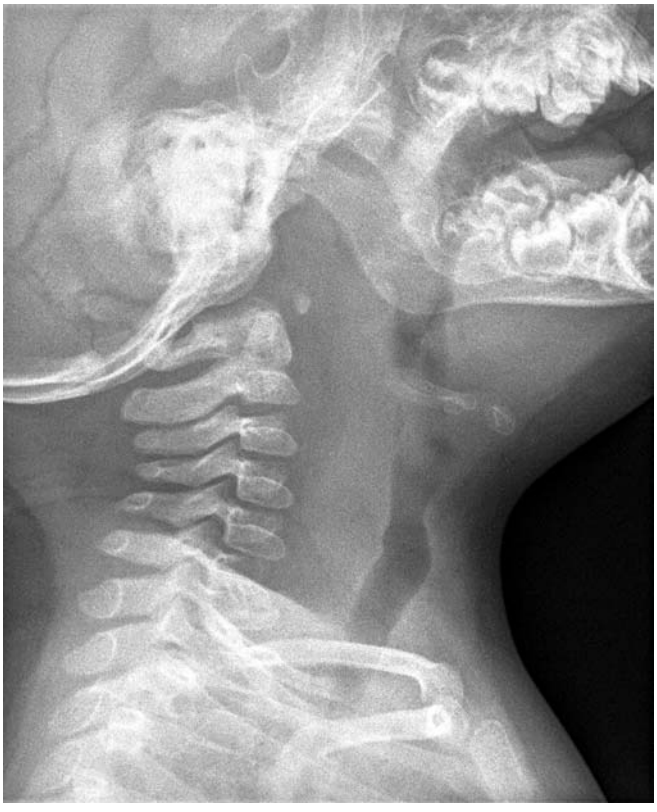

Fig. 4 An example of a MUSICA-processed pediatric image (Speed Class 400). (Image courtesy of Prof. G. Marchal, University Hospital Leuven, Belgium)

Other dose-management considerations

Digital imaging systems provide many advantages over their analogue counterparts, but their complexity also

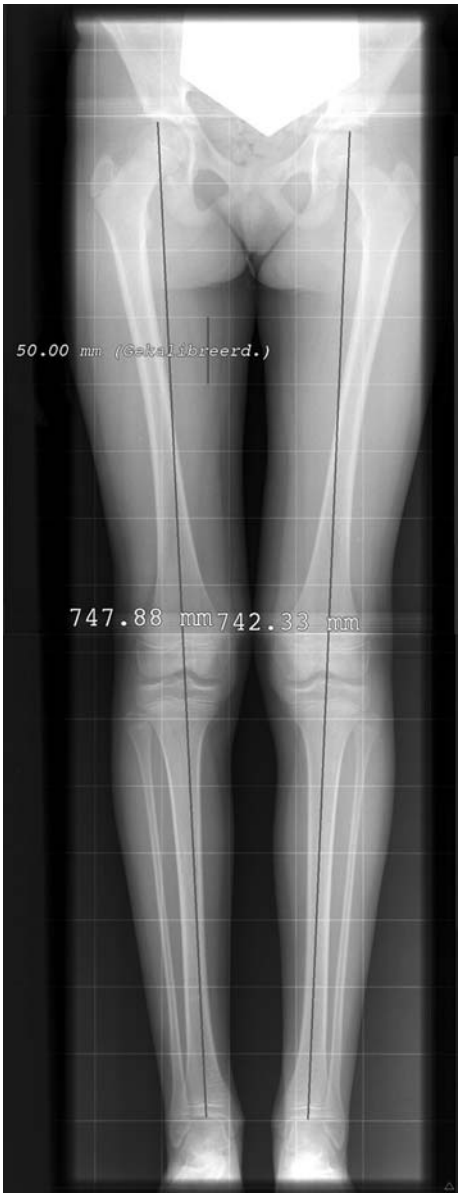

Fig. 5 An example of the Agfa ADC full-leg/full-spine software that stitches together individual images of a simultaneously exposed set to produce a single output image of body parts too long to fit on a single IP. (Speed Class 200). (Image courtesy of Prof. G. Marchal, University Hospital Leuven, Belgium)

creates additional burdens. One of these necessary burdens is quality control [18]. Without calibrated x-ray sources and well-characterized acquisition and display systems, dose management would be impossible, even pointless. Agfa provides test objects, procedures, and special, automated QC software for users to maintain their CR equipment at peak performance. There are tools to check, among other things, kV accuracy and collimation of the x-ray source, CR system sensitivity and SNR, contrast performance, geometrical fidelity, and image sharpness. Periodic QC using these tools will ensure that dose feedback and ongoing dose monitoring yield meaningful, actionable results.

A second QC burden is reject analysis. Currently, there is no provision on Agfa's acquisition workstations to do reject analysis (this is, however, a work in progress). Instead, some users are combining the reject results from multiple CR systems on the attached PACS to do the analysis. This has the advantage that results from all scanners are collected in a single, central database, rather than distributed across multiple machines. On the other hand, it requires a certain level of discipline amongst the radiographers to ensure that all rejected images are transferred to the PACS [19].

One concern about the easy transfer of image, patient and exam information in a PACS environment is the ease in altering some of that information. This is important for CR dose management, as well. Since some of the dose information, like kV, mAs, grids, filters, and SID is entered manually (when it is entered at all) and stored in the image's DICOM header, it is also susceptible to editing. Agfa's workstation allows users to edit this information, since there is the potential for data entry errors at acquisition that should be corrected before the image is released to the PACS. Proper security procedures must be in place on the PACS to prevent further unauthorized alterations. Some dose-related information generated within the Agfa CR system (like speed class, and lgM) is not alterable, and so provides a trail back to the original exposure, even if other details have been changed.

Finally, Agfa also provides a way to handle some of the special procedures that arise in pediatric imaging. In particular, the ADC full-leg/full-spine software provides a way to produce composite images from multiple, simultaneously exposed individual images that have been automatically stitched together. This is very useful in situations where the body part does not fit completely on one cassette, like scoliosis and full-leg exams. The IPs can be placed into a full-body cassette, as in S/F, or they can reside in smaller cassettes that are placed in a special holder designed for such exams. The holder contains a faint, coarse radiopaque grid. The x-ray projection of this registration grid generates a cross-hatch pattern in the image that is used by the software for geometrical alignment and linearity adjustments when stitching the separate images together (Fig. 5). Measurement and annotation tools at the workstation simplify subsequent diagnostic and planning tasks.

Summary

The increased focus on radiation dose and its management, particularly in the pediatric environment, has spurred the demand for new imaging procedures, better detectors, and improved hardware and software. Some of these improvements are quite similar across different manufacturers, while others are unique. Agfa provides special tools and techniques at all relevant levels of the imaging chain (exposure, acquisition, image processing, image and information management) that make the task of managing pediatric dose easier for users. New developments, like needle-based storage phosphors, next generation image-processing software, special equipment and software for pediatric applications, and new image management and QC tools will improve the situation further. While manufacturers clearly play an important role, dose management is a cooperative effort between manufacturers, users, and regulators. Only with such broad-based cooperation can we achieve the goals of ALARA in pediatric imaging.

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