Acceptance Testing and Quality Control of Photostimulable Storage Phosphor Imaging Systems

Report of AAPM Task Group 10

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Acceptance Testing and Quality Control of Photostimulable Storage Phosphor Imaging Systems

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ABSTRACT

Photostimulable phosphor (PSP) imaging, also commonly known as computed radiography (CR), employs reusable imaging plates and associated hardware and software to acquire and to display digital projection radiographs. Procedures to guide the diagnostic radiological physicist in the evaluation and continuous quality improvement of PSP imaging practice are the specific goal of this task group report. This document includes an overview of a typical PSP imaging system, functional specifications, testing methodology, and a bibliography. The main body of the report includes a description of generic, non-invasive tests that are applicable to a variety of PSP units. Since the inception of the task group, technological advances have changed the scope of its original goals. In particular, when the task group was initially formed, film was the chief display medium, and video display monitors and associated analysis software were not widely available. Also noted are the significant advances of "direct radiography," which includes the direct acquisition and readout of projection radiographs without physical handling of a PSP imaging plate. The user of this information should realize that technological change is constant, and many of the tests described in this document might not be applicable with the current state of the art. In these situations, the user is encouraged to consult the manufacturer, the American Association of Physicists in Medicine (AAPM), or other pertinent contemporary sources for additional information.

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1. INTRODUCTION

The primary purpose of this document is to guide the clinical medical physicist in the acceptance testing of photostimulable phosphor (PSP) imaging systems. PSP imaging devices (modalities) are known by a number of names including (most commonly) *computed radiography (CR), storage phosphor imaging, photostimulable storage phosphor imaging, digital storage phosphor imaging,* and *digital luminescence radiography.* PSP images are readily integrated into a picture archiving and communication system (PACS) via widely implemented communications standards including Digital Imaging and Communications in Medicine (DICOM)[1] and Health Level 7 (HL7).[2] The tests described herein are appropriate for PSP systems in either integrated or stand-alone applications. Digital imaging technology is rapidly evolving, and this effort represents the state of technology as of its writing. Proper application of this guide involves supplementing it with current literature and specific manufacturer's technical data.

A secondary purpose is to provide a consolidated source of information regarding device functionality, testing, and clinical practice of PSP imaging. This document provides the physicist with a means to conduct initial acceptance testing, to interpret results, and to establish baseline performance. A subset of these tests can be extended to routine quality control (QC).

2. SYSTEM OVERVIEW

The basic principles and operating characteristics of PSP systems are covered in this section, including acquisition methods, PSP detector characteristics, the readout process, and the detector characteristic response. Detailed information on the physics of computed radiography is available from the review article by Rowlands.[3]

2.1 PSP Image Acquisition

The ***photostimulable phosphor (PSP)*** stores absorbed x-ray energy in crystal structure "traps," and is sometimes referred to as a "storage" phosphor. This trapped energy can be released if stimulated by additional light energy of the proper wavelength by the process of ***photostimulated luminescence (PSL).*** Acquisition and display of the PSP image can be considered in five generalized steps, illustrated in Figure 1.

The unexposed PSP detector, commonly known as an imaging plate (IP), is placed in a cassette with a similar form factor and appearance of a screen-film cassette. X-ray geometry and imaging techniques are also similar to screen-film acquisition. During the exposure, x-rays are transmitted through the patient and are absorbed by the IP. Energy deposited in the PSP material causes local electrons to be elevated from an equilibrium (ground state) energy level to a stable "trap" known as an "F-center." This is the unobservable "electronic" latent image, whereby the number of electrons trapped is proportional to the number of x-ray photons incident on the IP. The exposed IP in step 1 of Figure 1 must be read out to produce the x-ray image. In step 2, the cassette is placed in the reader where the IP is extracted and raster-scanned with a highly focused and intense laser light of low energy (~2 eV). Trapped electrons in the PSP matrix are stimulated by the laser energy, and a significant fraction return to the lowest energy level within the phosphor, with a simultaneous release of PSL of higher energy (~3 eV). The intensity of PSL, proportional to the number of released electrons, is optically filtered from the laser light and captured by a light guide assembly in close proximity to the IP. A photomultiplier tube (PMT) at the light guide output converts and amplifies the PSL into a corresponding output voltage.
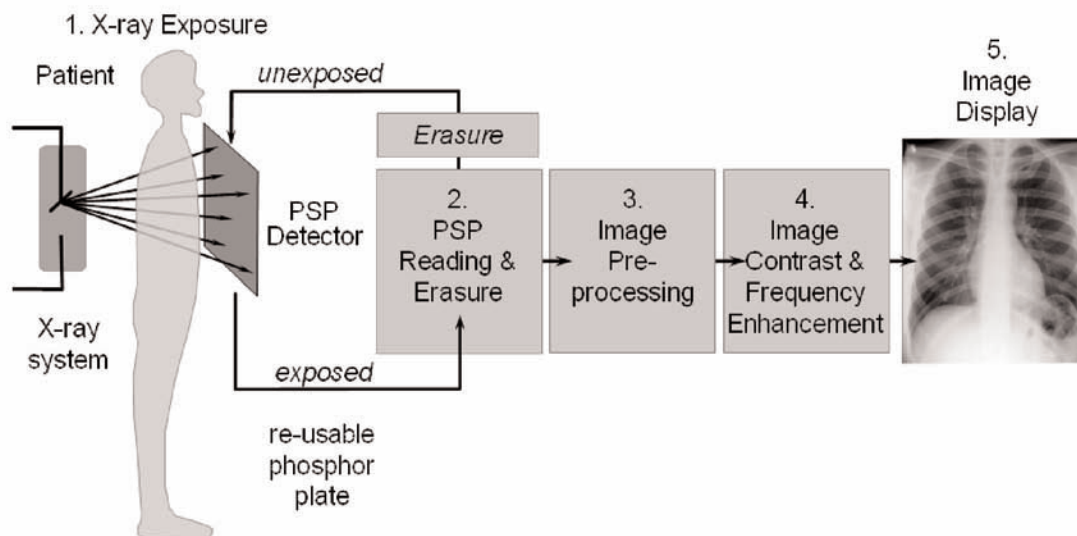
Figure 1. PSP Image acquisition and processing can be divided into five separate steps: (1) Image acquisition involves exposing the patient with a study-specific x-ray technique and recording the transmitted x-ray flux with the PSP detector. (2) The resultant latent image is extracted via the reader device using laser stimulation and recording the PSL intensity. (3) Image pre-processing involves correcting systematic variations in the extraction process and determining the range of pertinent information with subsequent adjustment of digital values to a normalized output range. (4) Image post-processing translates the digital values of the raw digital image to render a grayscale and frequency enhancement appropriate for the anatomy and study. (5) The output image is displayed on a calibrated image monitor for presentation.

Subsequent digitization using an analog-to-digital converter (ADC) produces a corresponding digital number at a specific location in the digital image matrix determined by the synchronization of the laser beam and IP location. Residual latent image information is erased using an intense light (consisting of wavelengths that remove electrons in traps without stimulating further electron trapping), and the IP is reinserted into the cassette for reuse. Image pre-processing takes place in step 3, to correct for static light guide sensitivity variations and fixed noise patterns, so that the imaged object is faithfully reproduced and scaled to a normalized range as "raw" image data. Wide dynamic range response of the PSP detector requires image recognition, scaling, and contrast enhancement to optimize the image characteristics and signal-to-noise ratio (SNR) of the "processed" image data in step 4. Display of the digital image in step 5 uses look-up-table (LUT) transformations to properly render the digital image code values into corresponding grayscale brightness variations for soft-copy monitors and optical density (OD) values for hard-copy film. In terms of acquisition, the PSP system closely emulates the conventional screen-film detector paradigm. There are, however, several important differences relative to screen-film detectors to realize the full advantage of PSP imaging capabilities, including collimation and position of the object on the detector, variable (selectable) detector speed, sensitivity to x-ray scatter, importance of optimal image processing, and image artifacts, among other issues.

Comparison to other digital radiography devices: Digital (direct) radiography (DR) devices are based upon the detection of the transmitted x-ray fluence through a patient using an "active" large field-of-view (FOV) detector, in which the x-ray energy is captured and converted to a latent image in the form of locally deposited charge. The latent image is converted directly to a digital image dataset without further system interaction by the operator. These devices have advantages in quick turnaround for image display; however, the systems are rather expensive,

most are not portable, and they sometimes do not have the flexibility to achieve appropriate patient positioning for views that are relatively easy for screen-film and PSP detectors.[4,5] Nevertheless, digital imaging devices are becoming less functionally distinct, as there are "cassetteless" PSP systems that emulate DR productivity, and a DR detector that demonstrates portable imaging capability usually associated with PSP systems. Similar function and common characteristics of PSP and DR systems allow the application of many procedures described in this document for acceptance testing and quality control for DR systems. There are also attributes for any digital system that require manufacturer-specific tests and criteria to determine and verify optimal operation.

2.2 PSP Detector Characteristics

PSP detectors are based on the principle of *photostimulated luminescence.*[6–9] When an x-ray photon deposits energy in a PSP material, three different physical processes account for energy conversion. *Fluorescence* is the prompt release of energy in the form of light. This process is the basis of conventional radiographic intensification screens. IPs also emit fluorescence in sufficient quantity to expose conventional radiographic film.[10,11] This, however, is not the intended method of imaging. PSP materials store a significant fraction of the deposited energy in crystal structure defects, thus the synonym *storage phosphors.* This stored energy constitutes the latent image. Over time, the latent image fades spontaneously by the process of *phosphorescence.* If stimulated by light of the proper wavelength, the process of *stimulated luminescence* can release a portion of the trapped energy immediately. The emitted light constitutes the signal for creating the digital image.

Many compounds possess the property of PSL; however, few have characteristics desirable for radiography, namely a stimulation-absorption peak at a wavelength produced by common lasers, a stimulated emission peak readily absorbed by common PMT input phosphors, and retention of the latent image without significant signal loss due to phosphorescence.[12] The compounds that most closely meet these requirements are alkali-earth halides. Commercial products (as of 2004) have been introduced based on $RbCl$, $BaFBr:Eu^{2+}$, $BaF(BrI):Eu^{2+}$, $BaFI:Eu^{2+}$, and $BaSrFBr:Eu^{2+}$.* A typical PSP detector is layered on an opaque substrate, as illustrated in Figure 2A. A PSP detector with an optically transparent base allowing extraction of the PSL light from both sides when stimulated is now clinically available[13] as shown in Figure 2B, and a structured phosphor is under investigation, comprising $CsBr$ (cesium bromine)[14,15] as artistically illustrated in Figure 2C. These latter two implementations show great promise in improving detection efficiency and image information transfer, resulting from improved detection efficiency and conversion efficiency.[3] In general, the PSP compounds, their formulation, and structural characteristics are tuned for a given manufacturer and often function optimally only with a specific reader; imaging plates are generally not interchangeable between readers.

2.2.1 Doping and Absorption Process

Trace amounts of Eu^{2+} impurities are added to the PSP to alter its structure and physical properties. The trace impurity, also called an *activator,* replaces the alkali earth in the crystal, forming a luminescence center. Ionization by absorption of x-rays (or ultraviolet [UV] radiation) forms electron/hole pairs in the PSP crystal. An electron/hole pair raises Eu^{2+} to an excited state, Eu^{3+}.

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*Ba: barium; Br: bromine; Cl: chlorine; Eu: europium; F: fluorine; I: iodine; Rb: rubidium; Sr: strontium
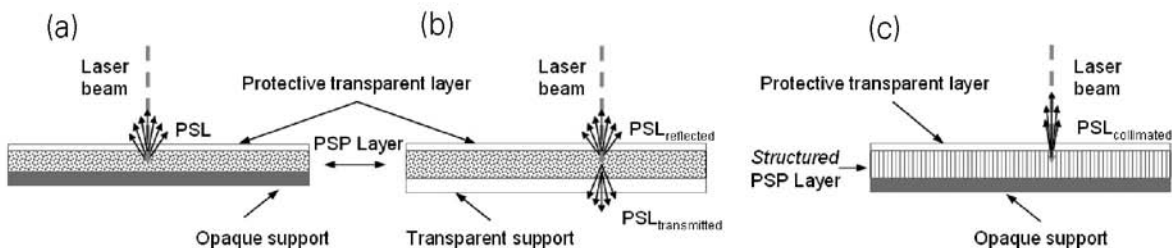
Figure 2. Cross-sectional views of the (a) generic, (b) dual-side readout, and (c) structured PSP detectors. Often, the opaque support will have a reflective layer to increase the PSL intensity.

Eu^{3+} produces visible light when it returns to the ground state, Eu^{2+}. Stored energy (in the form of trapped electrons) forms the latent image. There are currently two major theories for the PSP energy absorption process and subsequent formation of luminescence centers. These include a bimolecular recombination model[8] and a photostimulated luminescence complex (PSLC) model[9] shown in Figure 3. Physical processes occurring in $BaFBr:Eu^{2+}$ using the latter model appears to closely approximate the experimental findings. In this model, the PSLC is a metastable complex at higher energy ("F-center") in close proximity to an Eu^{3+}-Eu^{2+} recombination center. X-rays absorbed in the PSP induce the formation of "holes" and "electrons," which activate an "inactive PSLC" by being captured by an F-center, or form an active PSLC by formation and/or recombination of "exitons" explained by "F-center physics".[9] In either theoretical description, the numbers of active PSLCs created (number of electrons trapped in the metastable site) are *proportional to the x-rays locally absorbed in the phosphor.*

X-ray absorption efficiency of BaFBr:Eu is compared to $Gd_2O_2S:Tb$ (rare-earth screens) for typical thicknesses of material encountered, as shown by attenuation curves illustrated in Figure 4. Between 35 to 50 keV, the BaFBr phosphor has higher x-ray absorption per unit mass thickness due to the lower K-edge absorption of barium; however, below and above this range, the gadolinium rare-earth phosphor is superior. A typical incident x-ray beam on the conventional PSP often requires greater exposure to achieve similar image quantum statistics compared to a 400-speed rare-earth detector. In addition, the high absorption probability of x-rays below
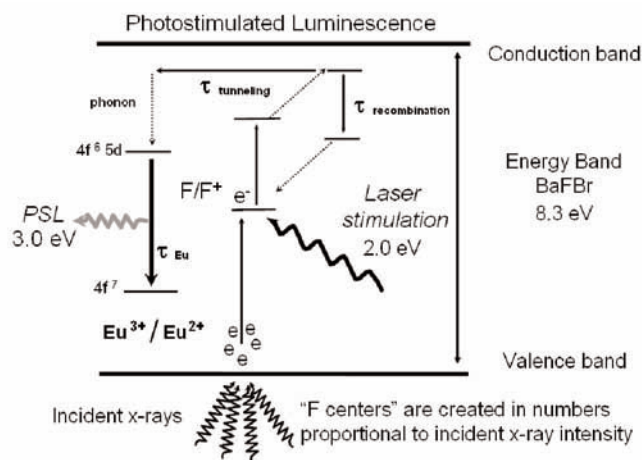
Figure 3. An energy diagram of the excitation and PSL processes in a $BaFBr:Eu^{2+}$ phosphor (von Seggern et al.[9]). Incident x-rays form an "electron" latent image in a metastable "F" center site that can be stimulated with a low-energy laser beam (~2.0 eV), producing the desired luminescent signals (~3.0 eV). τ is a constant representing the decay times for the specific indicated processes discussed in the text.

X-ray absorption Efficiency: BaFBr, CsI,Gd$_2$O$_2$S
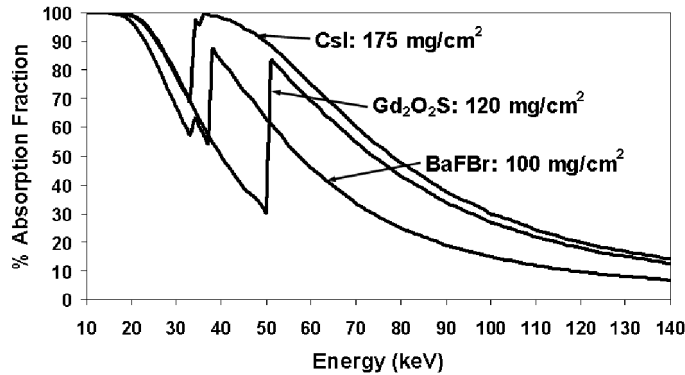
Figure 4. The photon absorption fractions of PSP and rare-earth x-ray phosphors are plotted as a function of energy. Phosphor thicknesses are representative of a standard 400-speed conventional screen, a "standard resolution" PSP detector (100 mg/cm^2), and a CsI phosphor commonly used in indirect thin-film transistor (TFT) array and optically coupled, charge-coupled device (CCD) camera DR systems.

50 keV, where a significant fraction of lower energy scattered x-rays occur, results in a greater sensitivity to scatter compared to the rare-earth absorber. The PSP detector is often mentioned as a "scatter sponge" in this context.[16]

2.2.2 Fading

Fading of the trapped signal will occur exponentially over time, through spontaneous *phosphorescence*. A typical imaging plate will lose about 25% of the stored signal between 10 minutes to 8 hours after an exposure, and more slowly afterwards.[17] Fading introduces uncertainties in output signal that can be controlled by introducing a fixed delay between exposure and readout to allow decay of the "prompt" phosphorescence of the stored signal. After about 10 minutes, the latent image fades more slowly.

2.2.3 Stimulation and Emission

The "electronic" latent image imprinted on the exposed BaFBr:Eu phosphor corresponds to the activated PLSCs (F-centers), whose local population of electrons is directly proportional to the incident x-ray fluence for a wide range of exposures, typically exceeding 10,000 to 1 (four orders of exposure magnitude). Stimulation of the Eu^{3+}-F-center complex and release of the stored electrons requires a minimum energy of ~2 eV, most easily deposited by a highly focused laser light source of a specific wavelength. HeNe (helium-neon, $\lambda=633$ nm) and "diode" ($\lambda\cong680$ nm) laser sources are most often used, with the latter becoming much more prominent. The incident laser energy excites electrons in the local F-center sites of the phosphor. According to von Seggern,[9] two subsequent energy pathways within the phosphor matrix are possible: (1) to return to the F-center site without escape, or (2) to "tunnel" to an adjacent Eu^{3+} complex. The latter event is more probable, where the electron cascades to an intermediate energy state with the release of a non-light-emitting "phonon" (mechanical energy release). A light photon of 3 eV ($\lambda\cong410$ nm) immediately follows as the electron drops through the energy level of the Eu^{3+} complex to the more stable Eu^{2+} energy level. Figure 5 shows a plot of the energy spectra of the laser-induced electron stimulation and subsequent light emission. Different phosphor formulations have opti-

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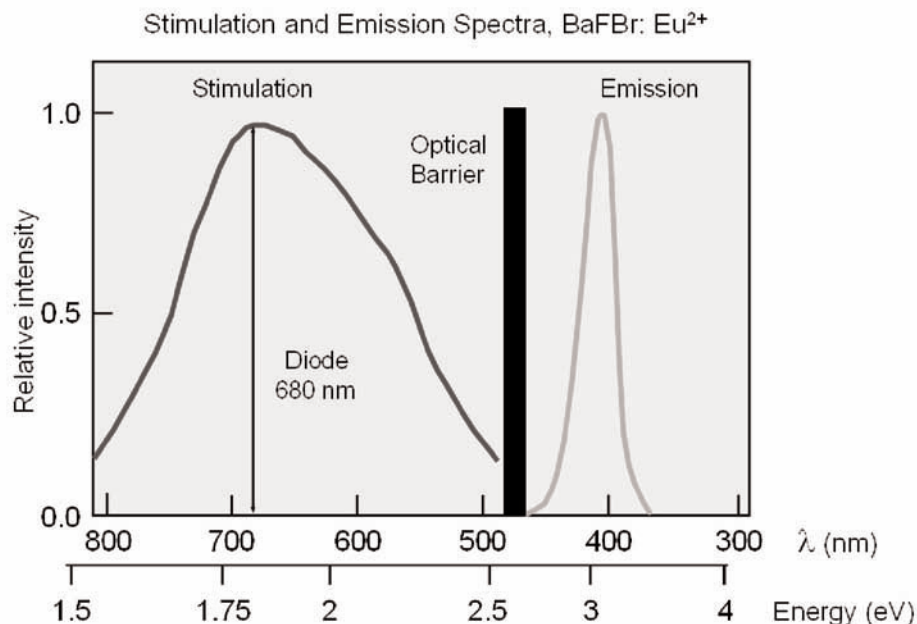
Figure 5. Stimulation and emission spectra for the BaFBr:Eu^{2+} storage phosphor demonstrates the energy separation of the excitation and emission events. Selective optical filtration isolates the light emission intensity from the incident laser intensity. In absolute terms, intensity of the emitted (PSL) light is significantly lower. Other PSP formulations exhibit similar spectra. (Figure adapted from reference 9 with permission from the American Institute of Physics.)

mal stimulation energies tuned to specific laser energy. For best imaging performance, it is best to use the phosphors designed for a particular PSP reader system.

Conversion efficiency represents the amount of energy that is extracted from the laser stimulation process as stimulated luminescence and the fraction of light that is captured and converted into a useful output signal. This is dependent on several factors, including the dwell time of the incident laser beam, the spot size of the laser beam, the depth of penetration of the laser beam into the phosphor, the amount of scattering of the PSL photons, the capture efficiency of the collecting light guide, the conversion efficiency of the light to electronic charge device (typically a PMT, but also a CCD photodiode array in some systems), and the accuracy/efficiency of signal digitization as detailed by Rowlands.[3] Acquisition/readout technologies such as dual-side readout[13,18] and needle-structured PSP phosphors[14,15] improve the conversion efficiency significantly, thus enhancing the overall statistical integrity and reducing the noise of the captured signal.

2.3 The Readout Process

The PSP reader and basic components are illustrated in Figure 6.

2.3.1 Point-scan Laser Readout

Produced by either a HeNe or diode laser source, a focused laser beam is routed through several optical components prior to scanning the phosphor plate. (Note: Most current systems since about the year 2000 use a diode laser.) A beam splitter uses a portion of the laser output to monitor the incident laser intensity via a reference detector, and to compensate the output PSL signal
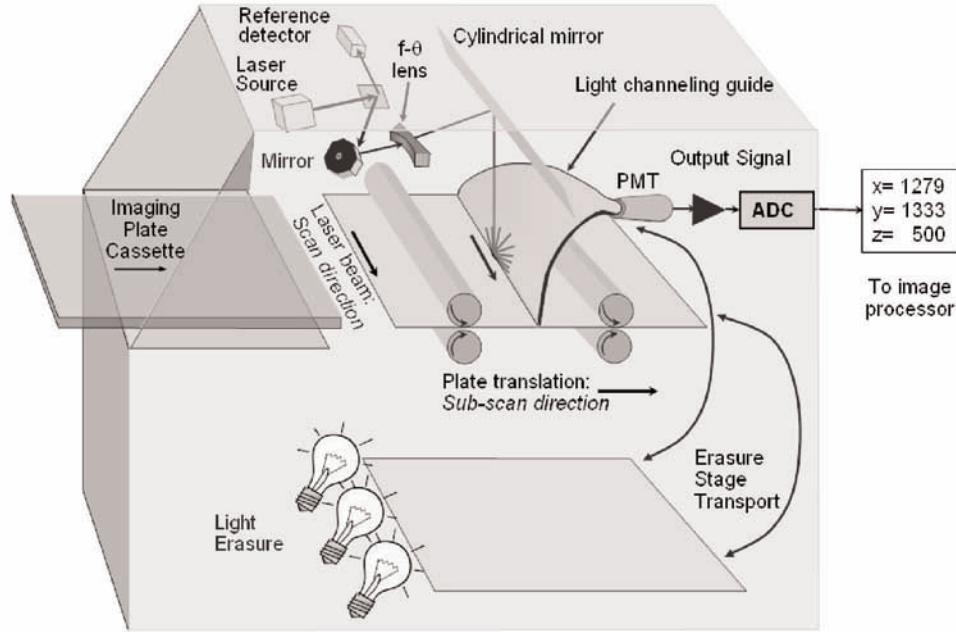
Figure 6. Major components of a PSP reader (point-scan, laser flying spot) include the stimulating laser source, a beam splitter, oscillating beam deflector, f-θ lens, cylindrical reflecting mirror, light collection guide, photo-multiplier tube (PMT), and light erasure stage. The IP is translated in a continuous motion through the laser beam scan by pinch rollers. All component functions are synchronized by digital computers. In some readers, multiple PMTs are used for capturing the signal. The erasure stage removes residual signals, and the IP is returned to the cassette.

intensity for incident power fluctuations.[19] The major portion of the laser energy reflects off a scanning device (rotating polygonal mirror or oscillating flat reflector), through an optical filter, shutter, and lens assembly. To maintain a constant focus and linear sweeping velocity across the PSP plate, the beam passes through an f-θ lens to a stationary mirror (typically a cylindrical and flat mirror combination). Assuming a Gaussian beam profile, the laser beam irradiance varies with radial distance r from the center, as $I(r) = I_0 \exp(-2r^2/r_1^2)$, where r_1 is the radial extent that the irradiance has dropped to $1/e^2$ of its value on the beam axis, I_0, (e = 2.71828…). This is a measure of the effective laser beam diameter. Typical laser "spot sizes" range from 50 μm to 200 μm and several sizes in between, depending on the manufacturer and reader as measured at the surface of the IP.

The speed of the laser beam across the phosphor plate is limited by the luminescent signal decay time constant (~0.7 to 0.8 μs for $BaFBr:Eu^{2+}$) following excitation[3] to maintain spatial resolution. Laser beam power determines the fraction of the electrons released from the F-centers, the fraction of phosphorescent lag, and the amount of residual signal. Higher laser power can release more trapped electrons, but the trade-off is a loss of spatial resolution caused by increased laser penetration depth and wider spread of the stimulated light in the phosphor layer. Signal decay lag (afterglow) causes blur in the scan direction, and results in loss of high-frequency response near the Nyquist frequency. At the end of the scanned line, the laser beam is retraced to the start and repeated.

Translation of the IP through the optical stage occurs continuously at a speed to ensure an "effective" sample size is equal in both the laser scan and plate sub-scan dimensions. This imposes an upper limit to spatial resolution in both dimensions.[19] Scanning and plate translation continues in a raster fashion over the total phosphor area. *Scan direction, laser scan direction,*

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or *fast-scan direction* are equivalent terminology referring to the direction of the laser beam. *Slow-scan, sub-scan,* or *translation direction* refer to the phosphor plate travel direction. The typical *scanning* time is chiefly limited by the laser scan speed; for a 35×43 cm imaging plate, the time varies by manufacturer, reader type, and laser resolution. In general, a scan time range of ~30 to 60 seconds is specified by most manufacturers. Newer phosphor formulations with less signal decay lag (e.g., BaFI:Eu = ~0.6 μs)[3] allows a faster scan speed without loss of resolution in the laser scan direction. IP readout geometry for a point-scan PSP reader is shown in Figure 7.

2.3.2 Dual-side Laser Readout

In 2001, a "dual-side" IP was introduced to acquire both reflected and transmitted PSL from a stimulating point laser source, with two light guides positioned on either side of the detector (see Figure 2). In this configuration, a larger fraction of stimulated light is captured, and with optimized frequency weighting of the reflected and transmitted signals,[13,18] a higher SNR is achievable than with the conventional single-sided readout, and good spatial resolution is maintained. As for comparison of detective quantum efficiency, enhancement of 40% to 50% has been reported,[20,21] which ultimately leads to improved dose efficiency and equivalent radiographic speed, or better SNR (as a selectable trade-off). Dual-side readout was initially produced for a prototype digital mammography detector, but is now being used for conventional PSP applications on readers with two light guides, in conjunction with the transparent base IP.

2.3.3 Line-scan Laser Readout

PSP systems based upon a laser line source coupled to an array of CCD photosensors were first clinically introduced in late 2003. These systems can read the latent image on a PSP plate in 5 to 10 seconds for a large FOV (35×43 cm) detector.[22,23] The schematic diagram in Figure 8
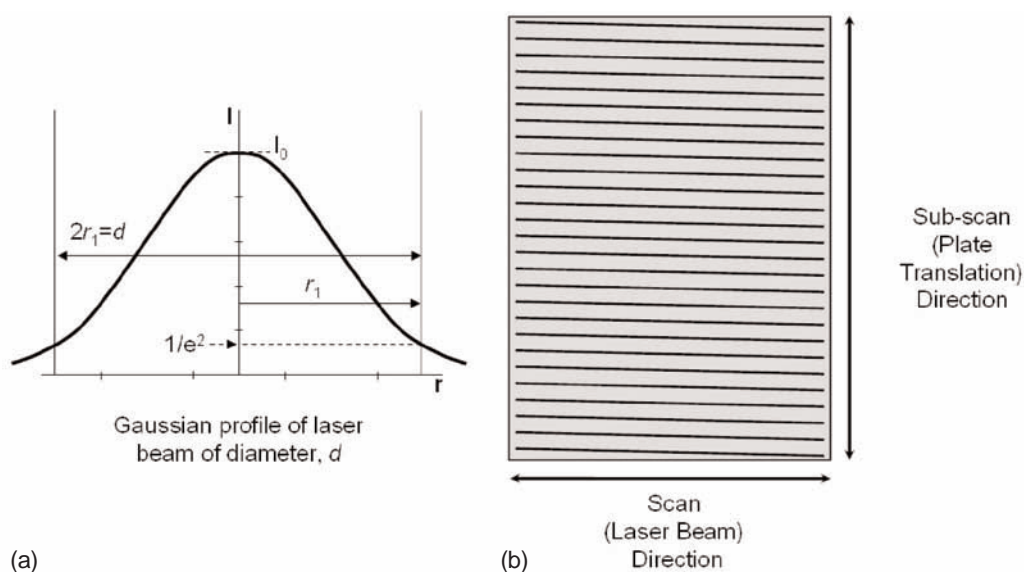
Figure 7. (a) Beam profile of laser of diameter *d*. (b) Diagram of the raster-scan of the phosphor detector indicates the fast-scan (laser scan) direction and the sub-scan (plate translation) direction. Note the slightly skewed angle of the readout lines relative to the edge of the phosphor plate, due to the simultaneous laser beam scanning and plate translation.
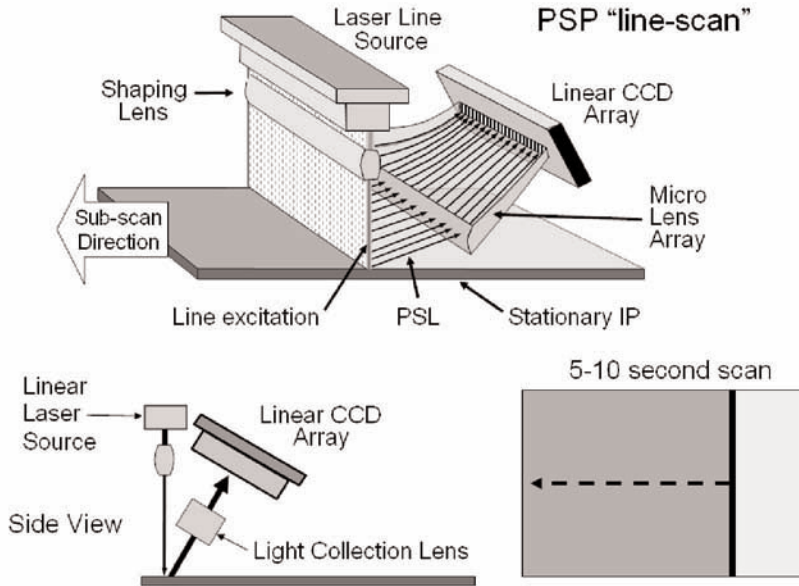
Figure 8. Schematic diagram of a "line-scan" PSP system, top, showing the general configuration, and bottom illustrating the side view of the components including the laser and light collection micro-lens array, and the geometry of acquisition. These systems are "cassetteless," and can read out the latent image in 5 to 10 seconds.

depicts the general configuration of a line-scan PSP system. Line excitation and readout of the IP reduces readout time by a large factor compared to a point scan system, without being limited by signal decay (phosphorescence) lag. A compact diode laser line source and micro-lenses to focus the PSL light photons onto the CCD photodiode array allow a small footprint and overall detector size. Line-scan PSP systems are competitive with DR devices in terms of processing speed, form-factor, and ease of use, with image performance similar to point-scan PSP systems.[3,15]

2.3.4 Residual Signal Erasure

Residual latent image signals are retained on the phosphor plate after readout. Residual signals are erased using a high-intensity light source of white or polychromatic content that flushes the traps without reintroducing electrons from the ground energy level. Unless an extreme overexposure occurs, essentially all of the residual trapped electrons are removed during the erase cycle. On most systems, the erasure of the plate is a function of the overall exposure, whereby higher incident exposures (e.g., in an uncollimated region of the detector) require a longer erasure cycle to eliminate possible image ghosting on subsequent uses of the imaging plate. In readers without "pipeline" processing (reading one IP while erasing the previous IP), the erasure time for highly exposed imaging plates can be a factor of 2 to 3 times longer than reading time, and a potential bottleneck for cycle time and throughput. A summary of the PSP detector cycle is illustrated in Figure 9.

2.3.5 Detection and Conversion of the PSL Signal

PSL is emitted in all directions from the phosphor screen for an unstructured PSP material. For a point-scan system, an optical collection system (mirror cavity or acrylic light collecting guide positioned at the laser phosphor interface along the scan direction) captures a portion of the emitted light, and channels it to the photocathode of the PMT (or PMTs) of the reader assembly. Detection
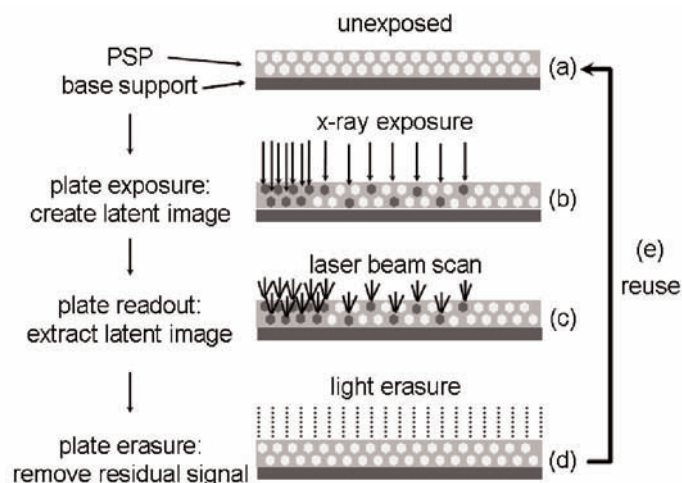
Figure 9. The phosphor plate cycle is depicted above. (a) An unexposed plate comprises the PSP material layered on a base support and protected by a thin, transparent coating. (b) Exposure to x-rays creates latent image centers of electrons semistable energy traps in the crystal structure. (c) Latent image processing is accomplished with a raster-scanned laser beam. Trapped electrons are released from the luminescent centers and produce light that is collected by a light guide assembly and directed to a PMT. (d) Residual trapped electrons are removed with a high-intensity light source, and (e) the plate is returned to the cassette for reuse.

sensitivity of the photocathode material is matched to the wavelength of the PSL (e.g., ~400 nm), and the light intensity releases a proportional number of photoelectrons. The electrons are accelerated and amplified by a cascading process through a series of dynodes within the PMT. Gain (and thus detector sensitivity) is internally adjusted by setting the dynode voltage to obtain a predefined target average output current for a given incident x-ray exposure typical of clinical images. (In some systems, the gain can be set by software controls.) An integration time for collecting the PSL signals is dependent on the laser scan speed and sampling pitch. Electronic dynamic range (minimum to maximum signal output) of the PMT is much greater than for the phosphor plate, and light intensity variations that correspond to incident radiation exposure respond linearly over a range of 10,000 or "four orders of magnitude" relative to the smallest *useful* signal.

Digitization of the output signal requires the identification of a minimum and maximum signal range, since most clinically relevant transmitted exposure variations occur over a dynamic range of 100 to 400. In early versions of PSP readers, a low energy laser pre-scan coarsely sampled the exposed PSP detector and determined the useful exposure range. The gain of the PMT was then adjusted (increased or decreased) to digitize the PSL over a predetermined intensity range during the high-energy laser scan. In most current systems, the PMT amplifier is preadjusted to be sensitive to the PSL resulting from an exposure range corresponding over a range of 2.58×10^{-9} C/kg (0.01 mR) to 2.58×10^{-5} C/kg (100 mR). This is equivalent to incident air kerma of 0.09 to 900 µGy. In addition, most manufacturers have the ability to set the sensitivity and the exposure dynamic range over a predefined range (for instance, a range for general radiography and a range for mammography).

In line-scan systems, the laser line source and the PSL emitter lens array localize the excitation and emission events on the IP,[23] focusing the output onto a CCD light-sensitive array for electronic conversion and digitization. CCD arrays do not have an exposure dynamic range as wide as a PMT, which can be a limiting factor with systems using this technology. Information on spatial resolution, dynamic range, amplification, and signal pre-processing of these systems is not yet available.

In all PSP systems, the output signal is nonlinearly amplified using a logarithmic or square root function. Logarithmic conversion provides an approximate linear relationship to x-ray attenuation and therefore to the transmitted x-ray intensity through an object, while square-root amplification provides a linear relationship with the quantum noise associated with the detected exposure.[24] In either case, lower signal intensities are expanded and higher signal intensities are compressed in the raw output signal. Many systems use nonlinear analog amplification of the output signal followed by digitization; some systems first digitize the output signal and use nonlinear digital methods to convert the signals. Digitization requirements are more severe for preserving quantitative integrity for the digital-to-digital conversion, with much greater bit depth needed to minimize quantization errors for the low amplitude signals.[25]

2.3.6 Digitization

Digitization is a two-step process of converting a continuous analog signal into a series of discrete digital values. The signal must be *sampled* and *quantized.* Sampling determines the location and size of the PSL signal from a specific area of the PSP detector, and quantizing determines the average value of the signal amplitude within the sample area. The output of the PMT is measured at a specific temporal frequency coordinated with the laser scan rate, and quantized to a discrete integer value dependent on the amplitude of the signal and the total number of possible digital values. The ADC converts the PMT signals at a rate corresponding to the number of pixels in the scan direction divided by the time per line. A pixel clock is synchronized to the absolute scan beam position and the corresponding position in the digital matrix. The translation speed of the phosphor plate in the sub-scan direction coordinates with the fast-scan pixel dimension so that the width of the line is equal to the length of the pixel (i.e., the pixels are "square"). The pixel pitch (distance between samples) is typically between 100 and 200 μm, depending on the dimensions of the IP, but may be as small as 50 μm for dedicated mammography systems. The sampling aperture is the area over which the signal information is averaged. This is determined by the laser beam distribution, and ideally is equal to the full-width-half-maximum (FWHM). Since the distribution has a Gaussian shape, the generated PSL signal extends beyond the pixel aperture, and the measured spatial resolution is usually less than what the pixel pitch and pixel aperture settings infer.

Although the analog output from the PMT has an infinite range of possible values between a minimum and maximum voltage, the ADC breaks the signal into a series of discrete integer values (analog-to-digital units or "code values") for encoding signal amplitude. The number of bits used to approximate the analog signal, or "pixel depth" determines the number of integer values. PSP systems typically have 10 to 16 bit ADCs, so there are $2^{10} = 1024$ to $2^{16} = 65,536$ possible code values for a given analog signal amplitude. One manufacturer uses a very large bit depth (16 bits or greater) to implement a *digital* logarithmic transformation to a final 12-bit/pixel image. Other system manufacturers use an *analog* logarithmic amplifier or a square-root amplifier on the *predigitized* signal. Analog amplification avoids quantization errors in the signal estimate when the number of ADC bits (quantization levels) is limited.[25]

2.3.7 Image Pre-Processing

Also known as "shading" or one-dimensional (1-D) "flat-fielding," the pre-processing algorithms reduce the variations in the sensitivity of the light collection guide.[26] Collection efficiency is lower at the edges of the IP because the light guides do not pick up as much light compared

to the center, as light diffuses in all directions. There also may be positionally dependent variations in the light collection guide sensitivity that cause low-frequency patterns. Each manufacturer has methods to measure and correct these nonuniformities. A basic correction scheme is illustrated in Figure 10A, indicating the measured profile, the "shading corrected" profile, and the resultant shading corrected image. Note that this is implemented in the fast-scan direction only. Dust or dirt on the light collection guide is easily visible as a linear artifact on the output image. These particles should be cleaned from the light guide surface, and not be present during a shading calibration, as this will lead to a different artifact in all subsequently processed images. A similar signal variation correction method is employed for line-scan PSP systems, but because the detector is in a fixed, unchanging geometry, a two-dimensional (2-D) flat field can be performed. This procedure can correct spatial variations in the translation direction as well. A 2-D flat-field correction matrix is created from a series of uniform exposures on the detector by averaging, normalizing the mean image, and inverting the response. Corrections are applied by taking the product of the flat-field image with the uncorrected raw image.[26] In both 1-D shading and 2-D flat-field corrections, the shading/flat-field matrices are also noisy (although the noise is reduced significantly with signal averaging). So, correcting a noisy patient image with a noisy correction image actually results in a greater output noise; however, the correction effectively removes the lower frequency spectrum variations and generally improves image quality.

2.4 Detector Characteristic Response

A linear, wide latitude response to variations in incident exposure is characteristic of the phosphor plate, while film is optimally sensitive to a restricted range of exposures. Figure 11 illustrates the characteristic curve response of a typical PSP detector to a 400-speed screen-film system. For screen-film detectors, which serve as both the acquisition *and* the display medium, it is necessary to tune the detector (film) contrast and radiographic speed to a narrow exposure range to achieve images with optimal contrast and minimal noise characteristics. PSP (and DR)
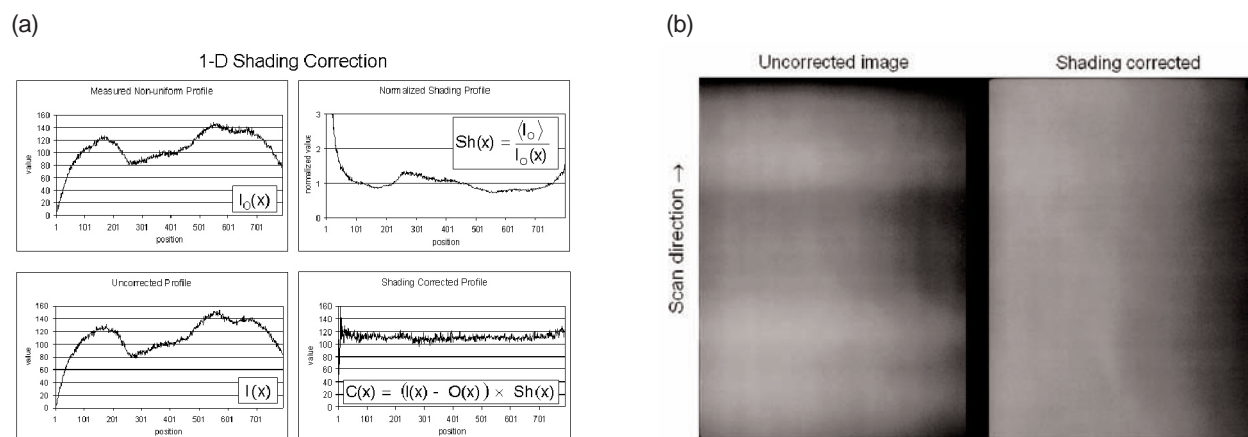
Figure 10. (a) The "shading correction" pre-processing step corrects for large and small area variations in light guide sensitivity performance by measuring the response of a known uniform field, producing a normalized, inverted response, and applying it to a specific image. (b) Typical pre and post "shading correction" processing shows the ability of the correction to reduce the nonuniformity in the scan direction, but not in the perpendicular direction (the plate translation direction), where lower intensity values on the right side of the image are due to the "heel effect" of the x-ray tube for a mammography application. (Reproduced from reference 26 by permission of the Radiological Society of North America (RSNA), Oak Brook, IL.)
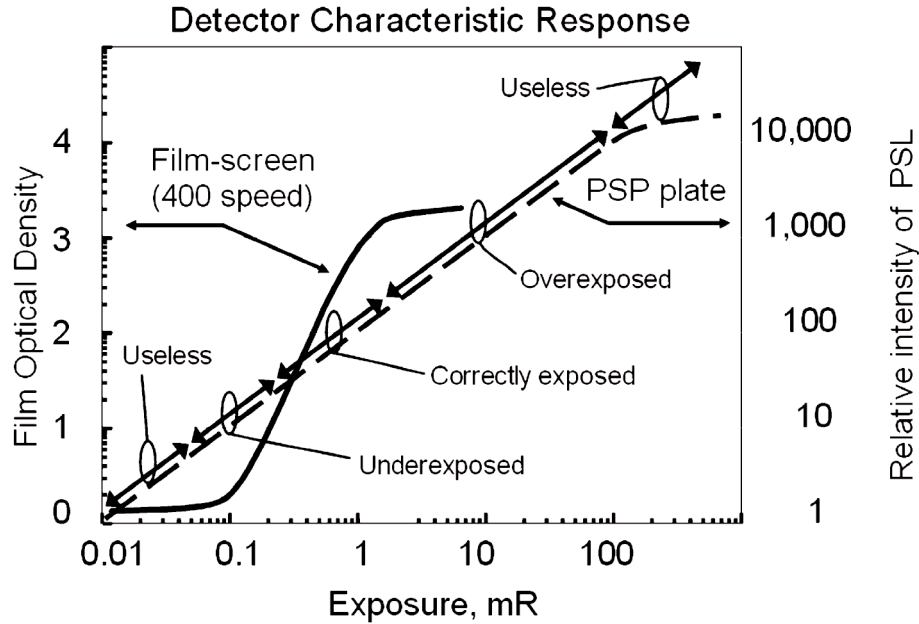
Figure 11. The characteristic response of a 400-speed rare-earth screen-film (solid S-shaped curve) and the PSP detector (dashed curve) are compared. Double arrows roughly indicate the exposure ranges characterized as underexposed, correct, or overexposed. "Useless" areas depict system responses that do not contain information useful for diagnosis either due to excessive quantum noise or saturation of the PSL mapped to digital number.

detectors are not constrained by the same requirements because the acquisition and display events occur separately, and compensation for under- and overexposures is possible by appropriate amplification of the digital data. However, identification of useful signal range must be accomplished prior to the auto-ranging and contrast enhancement of the output image. In addition, since under- or overexposed images can be "masked" by the system, a method to track exposures on an image-by-image basis is necessary to recognize those situations that exceed the desired or target exposure range so that action can be taken to resolve any problems. Of particular note is the broad range of over-exposure as shown in Figure 11, which can lead to "dose creep" (a subtle or gradual and potentially unnoticed increase in exposure when using digital detectors)[27] and excessive radiation dose to the patient. Exposure ranges marked "useless" represent average incident exposures that produce a significant fraction of signals over the image either so small as to be dominated by quantum noise, or so extreme as to be saturated. In either case, amplification adjustments cannot be made to extract any pertinent image information.

3. PROCESSING THE RAW PSP IMAGE

The characteristic response of the PSP detector has a slope of 1 over its dynamic range as shown by the log/log trace in Figure 11 (unlike the gradient of the screen-film response which has a slope typically greater than 3 over a very limited latitude). This wide latitude response translates to low radiographic (display) contrast if the full IP sensitivity range is matched to the luminance range of most medical displays. Identification of the pertinent information contained in the raw image is necessary so that only the useful signals relevant to clinical diagnosis are contrast enhanced, with the remainder ignored, otherwise the display dynamic range is occupied by unwanted signals and the image presentation will be suboptimal.

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3.1 Readout Parameters

3.1.1 Wanted vs. Unwanted Image Signals

In conventional screen-film radiography, the x-ray technologist adjusts the exposure technique to put the desired range of image signals on the linear portion of the H&D (Hurter & Driffield) curve. The image signals from x-rays outside of the object yet on the detector fall into the shoulder (high-exposure range) of the curve, and the image signals beyond the edges of the collimators fall into the toe (low-exposure range). The PSP system must similarly encode the useful image signal (values of interest, or VOIs), to provide maximum contrast sensitivity through look-up-table adjustments of the digital values (VOI LUTs). Just as the radiographic technique, collimation, and image detector cassette size are selected for the specific anatomic view, the PSP readout algorithms must make adjustments to the digital image specific to the anatomy.

3.1.2 Partitioned Pattern and Exposure Field Recognition

The first task is to determine the number and orientation of views in the raw digital data on the exposed detector (segmentation). While multiple views on a single cassette are good practice in conventional radiography, it is a potential complication for PSP radiography. Recommended is a single view per imaging plate in a PACS environment, although there are some vendors that can distinguish multiple images on a single IP and apply independent image processing and patient demographic information. Within an exposure field, it is important for the PSP reader to distinguish the useful region of the image by locating the collimation edges. Some PSP systems further segment the image by defining the edges of the anatomic region. Once the useful image content is correctly located, the PSP system disregards the image information beyond the collimator boundaries when performing further analysis.

3.1.3 Histogram Analysis

A method for determining the useful signal range for most PSP systems requires the construction of a *grayscale histogram* of the image, a graph of pixel value on the x-axis and frequency of occurrence on the y-axis (i.e., a spectrum of pixel values). The general shape of a histogram is dependent on the anatomy and the radiographic techniques employed for the image acquisition. Many PSP readers employ a histogram analysis algorithm to identify and classify the values that correspond to bone, soft tissue, skin, contrast media, collimation, unattenuated x-rays, and other signals. This allows the discrimination of the useful and unimportant areas of the image so that the image grayscale range can be applied to the anatomical information and properly rendered. An example of a chest-specific histogram is shown in Figure 12.

The result of histogram analysis allows the normalization of raw image data for standard conditions of speed, contrast, and latitude. Rescaling and contrast enhancements are optimized to render the appropriate image grayscale characteristics for the specific patient examination. Each manufacturer implements a specific method for this remapping procedure. With some systems, the latent image information is identified, logarithmically or square-root amplified, and resampled over a smaller range of digital values to minimize quantization errors. However, any errors in identification of the exposure range can be irreversible and require reacquisition of the image. Other systems digitize the full dynamic range of the PSL signal with a large ADC bit depth (e.g., greater than 16 bits) and then apply remapping algorithms to the digital data. In
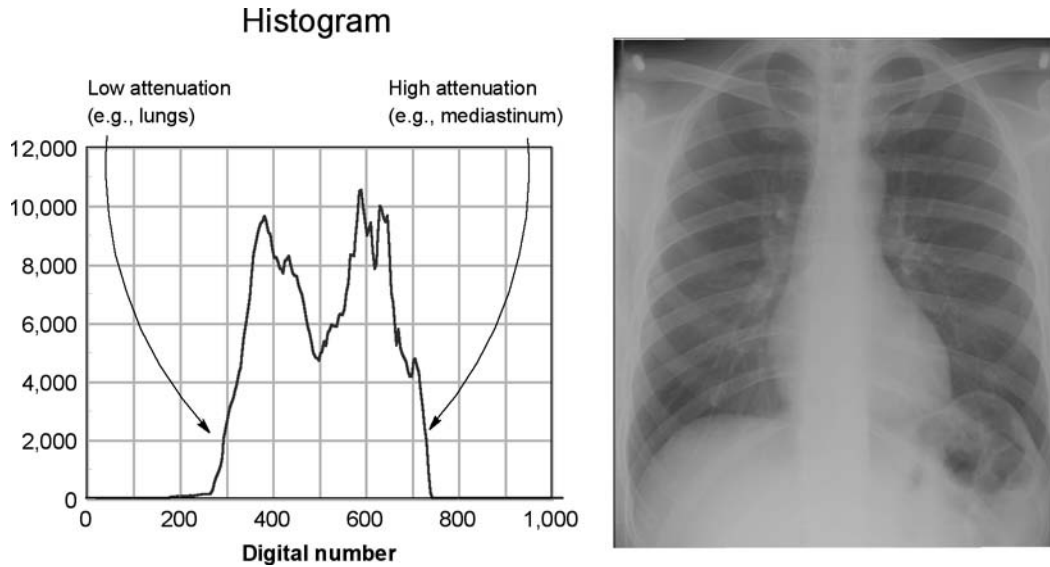
Figure 12. A chest histogram illustrates the various components of the frequency distribution of pixel values within the active area of the image, corresponding to anatomical variations. In this example, the digital values are directly related to the attenuation, similar to a screen-film image, by using a reverse LUT that inverts the digital representation of the collected PSL.

either case, the pertinent image information on the phosphor plate must be identified for subsequent grayscale and/or frequency processing, as the shape and information content of the histogram affects the processing of the image. An example of identifying and linearly amplifying the image signal, also known as *autoranging,* is described in Figure 13 for two exposure scenarios. In each case, the proper output range of digital values is obtained and produces "scaled" image data.

Because histogram shapes are anatomy and examination dependent, improper identification of the histogram minimum and maximum useful values can result in significant errors in image data scaling. For example, if one acquires a chest image but uses processing algorithms tuned for extremities, a potential misidentification of the proper histogram range can lead to unpredictable and suboptimal results.

3.2 Image Grayscale Adjustments

PSP images are matrices of digital pixel values that are readily manipulated to produce alternative image presentations. Three broad categories of processing include image contrast enhancement, spatial frequency modification, and other special image algorithm implementations.

To process images, PSP systems' manufacturers provide computer hardware and software, much of which is proprietary. Some third-party vendors provide similar functionality for remote processing of PSP image data. Selection and optimization of processing parameters is a nontrivial task that potentially requires "many thousands of man-hours by highly skilled staff."[28] A common problem is that the range of processing parameters far exceeds clinically useful values and can lead to gross "over-processing" artifacts. Modification of processing parameters must not be undertaken casually, and should be performed in conjunction with the application specialist with the consultation of the radiologists affected for optimization to the extent possible.

15
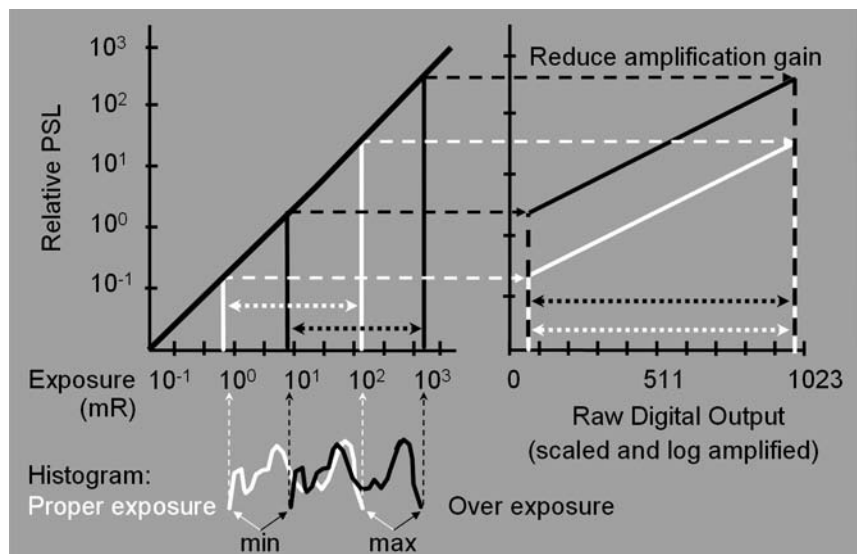
Figure 13. Autoranging of incident exposure into a corresponding digital number range is accomplished by analyzing the image histogram (lower left). Minimum and maximum values of the histogram (white curve, representing correct exposure) are mapped to minimum and maximum digital values using a 10-bit output LUT. Overexposure shifts the histogram distribution to a higher range (black curve), but the shape remains the same. The amplitude gain (digital or analog) is adjusted to map the pertinent data to a similar output range.

3.2.1 Contrast Processing

Because of small differences in attenuation of the human body and wide latitude of the PSP detector, there is very little inherent contrast in the raw image. To increase the visibility of anatomic detail, manufacturers provide *contrast-processing* software. The purpose of contrast processing is to create an image dataset with contrast similar to conventional screen-film images (at least as a starting point), or to enhance the conspicuity of desirable features. This type of processing is also referred to as *gradation processing, tone scaling,* and *contrast enhancement* by the various vendors.

There are several different methods employed for contrast processing. The most common and simple technique remaps individual pixel values according to system-applied LUTs to mimic the response of film contrast (Figure 14). A global modification of the contrast curve produces different local contrast from identical features at different grayscale levels. Each PSP system manufacturer has proprietary algorithms for applying contrast enhancement. Unacceptable image quality is often due to inappropriate processing even though the original "raw" image data is acquired properly.

A second type of contrast processing modifies contrast by performing operations on filtered versions of the original image and reconstructing an enhanced version of the original. There are several variants of "multiscale, multifrequency" processing, the details of which are obtainable from the respective manufacturer documentation.[28,30–32] These advanced processing algorithms provide an ability to reduce dynamic range and allow contrast and spatial frequency enhancement across the image.
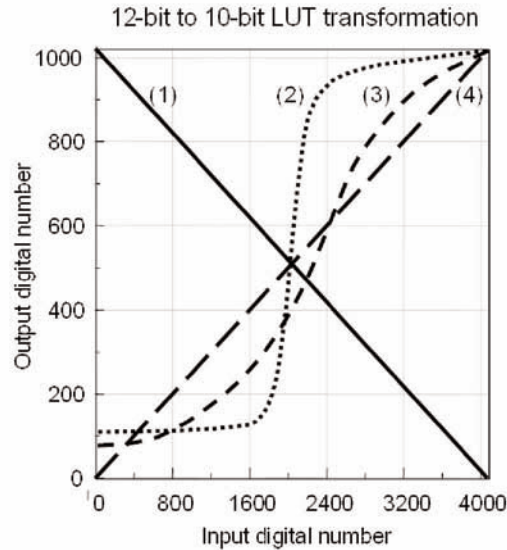
Figure 14. Grayscale conversion of the input (scaled, raw) digital values into output values occurs via a LUT transformation. Depicted above is a LUT that converts 12-bit input data into 10-bit output data. Image appearance (contrast) is also modified. Curve (1) inverts image contrast; curve (2) produces very narrow latitude and high-contrast output image; curve (3) has less contrast enhancement; curve (4) linearly passes the original data with unchanged contrast. (Adapted from a Fuji technical report, reference 29).

3.2.2 Frequency Processing

One purpose of digital image processing is to enhance the conspicuity of features within the image. *Frequency processing* enhances features characterized by specific spatial frequency content and weighting. Several techniques exist, the most common of which is blurred-mask subtraction.[33,34] Early users of PSP systems routinely printed an image twice on a single film using different presentations, one presentation designed to mimic the appearance of a conventional screen-film combination, and the other with significant amounts of "edge-enhancement." This practice is no longer routine, as image presentation is now most often performed with softcopy display.

Blurred-mask subtraction is a simple technique that blurs the original image with a convolution kernel of selected extent. Convolution is a process by which the kernel and the pixel values are multiplied by their corresponding values, added together, and then normalized by the summed value of the kernel. This value is the new central value of the "filtered" image. For example, a 3×3 kernel of all values equal to 1 has a normalization value of 9, and the product of the kernel values with the image are summed, divided by 9, and placed in the output image at the center pixel at same row and column position. The kernel is applied to each pixel location and surround in the original image to produce a low-pass filtered image. Characteristics of the kernel (weighting and extent) affect the characteristics of the frequency bandpass.[25] In the mask-subtraction process, the blurred image (of lower spatial frequency content) is subtracted from the original, producing a difference image containing predominantly high-frequency content. The difference image is multiplied by a constant to increase or decrease the bandpass amplitude, and then is added to the original image, producing the edge-enhanced image (Figure 15).

Multifrequency enhancement processing is now common, whereby the image is decomposed into distinct frequency ranges, typically achieved by using multiple convolution kernels of variable extent,[30] Laplacian pyramid decomposition,[31] wavelet decomposition techniques,[31] or

17
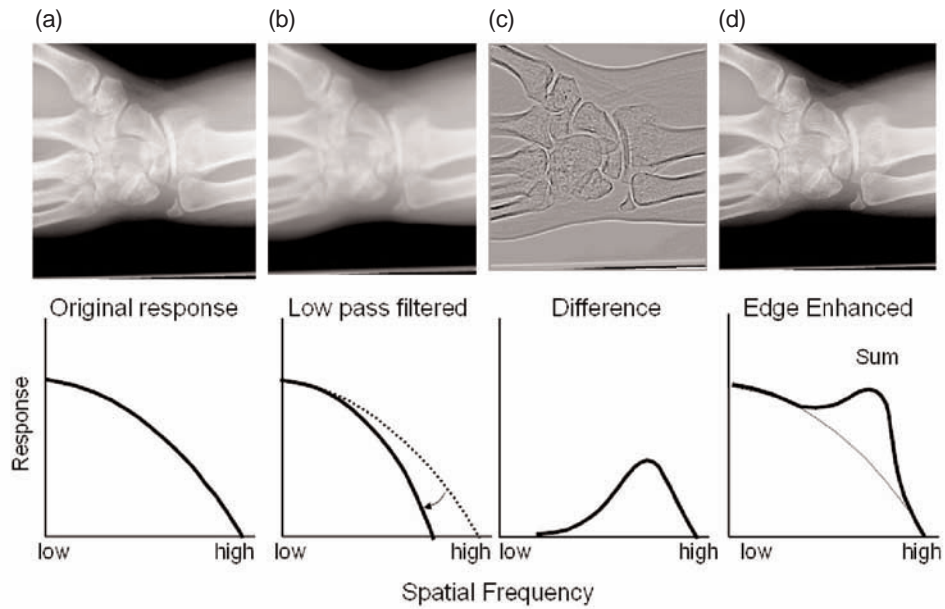
Figure 15. Edge enhancement example. (a) The original image frequency response (solid line) shows a monotonically decreasing signal modulation for higher frequency content. (b) A convolution filter blurs and reduces the high-frequency response (a low-pass filter). (c) The result of subtracting the blurred image from the original produces a bandpass response, dependent on the convolution kernel extent. (d) The scaled difference image is added back to the original and boosts the mid- to high-frequency range in the output (filtered) image.

other methods.[32] Each frequency bandpass range from the original image is independently weighted, normalized, and summed to reconstitute the final enhanced output image, extending both contrast and spatial resolution enhancement simultaneously across all image scales and gray levels. A simple example of the methodology is illustrated in Figure 16.

Generalized image grayscale enhancement and frequency processing examples of a PSP image are illustrated in Figure 17.
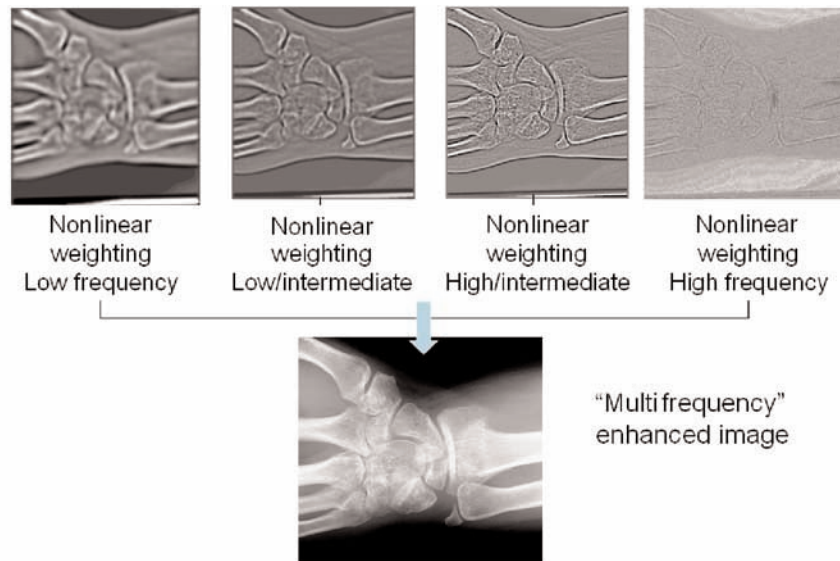
Figure 16. Multiscale (frequency) processing divides the image into frequency ranges that are independently processed and recombined to create the enhanced output image. Sub-images shown above are depicted with the same spatial extent, although in actuality they are of different "size."
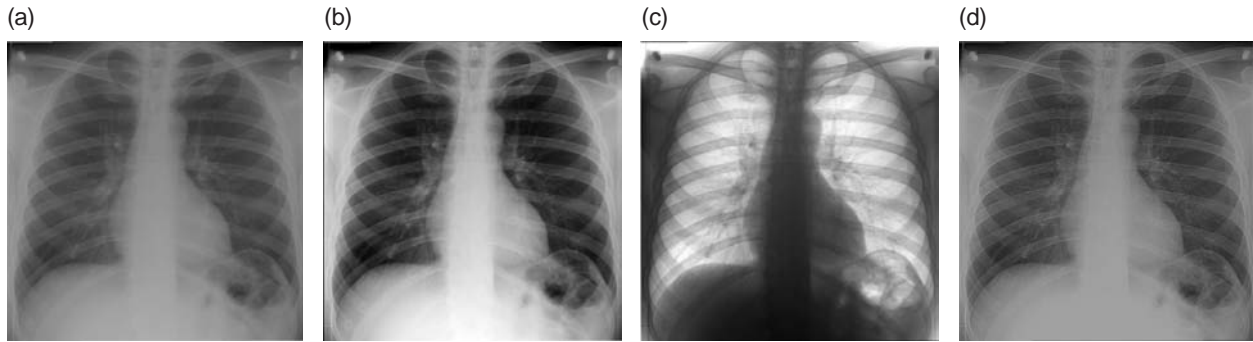
Figure 17. Examples of chest images demonstrating the flexibility of PSP systems and the variable contrast enhancement available. (a) Original "raw" chest image without contrast enhancement. (b) Contrast enhancement applied. (c) "Black-bone" or reversed contrast—often helpful in identifying tube placement. (d) Edge-enhanced image.

3.3 Other Image Processing.

Manufacturers have developed special processing software to address specific PSP applications. These include but are not limited to dual energy subtraction,[35–41] tomographic artifact suppression,[29] and scoliosis image acquisition and image stitching.[15,42,43]

"Disease-specific" image processing[44–46] to assist in the diagnosis of particular findings by making anatomic structures associated with disease more conspicuous by either enhancing the object or reducing the background are being introduced on all PSP and DR systems, typically as options. Computer-aided diagnosis/detection algorithms are used in conjunction with digital images to assist the radiologist. These algorithms depend on the quantitative accuracy of the data provided, and therefore require calibration of the PSP system.

4. IMAGE DEMOGRAPHICS AND EXPOSURE INDICATORS

4.1 Demographics and Processing Parameters

It is very important to understand and to be able to decode the information available on the hard-copy film or the soft-copy image. The manufacturer's user manual should contain pertinent information regarding pixel dimension, image magnification/reduction factors, type of LUTs, frequency enhancement settings, latitude of the image data, and incident exposure information, among other vendor-specific factors. For soft-copy images, information is available in the DICOM header, and should be mapped to the image overlay on the PACS workstation (a PACS-dependent feature). Re-sellers will "brand" their own demographics, notations, processing parameters, or limit features, even though a PSP system is manufactured by another company that may sell the identical hardware in a different configuration. The user should not assume that the information or capabilities are identical.

4.2 Exposure Indicators

The PSP system provides consistent OD or image grayscale output values for under- or overexposures on account of the wide latitude response and algorithms that scale the signal to a predetermined output range. More problematic are the overexposures, which can extend over a significant range (see Figure 11). Inappropriate radiographic techniques can easily be overlooked

or hidden. Therefore, an indicator of the average incident exposure on the IP is important to verify proper radiographic technique. Each PSP manufacturer has a specific method for providing this information. For instance, systems made by Fuji report a ***Sensitivity*** number, which is inversely proportional to the incident exposure. Kodak systems provide an ***Exposure Index,*** which is directly proportional to the logarithm of the exposure. Agfa products provide an indicator called ***lgM,*** whose value also varies in proportion to the logarithm of exposure. Konica systems provide an incident exposure indicator called ***REX*** (relative exposure). These (and other) relative exposure indicators are dependent on the energy absorbed in the detector and amplitude of PSL released during processing. Thus, energy dependence and phosphorescence decay (delay in processing) will have an influence on the exposure index response. Morphological segmentation and histogram analysis also affect the exposure indicator, and inappropriately applied processing algorithms can lead to similar variability in the reported exposure index.

For all digital systems, the exposure to the IP varies across the image when imaging an object, and it is impossible to represent the exposure by any single scalar value, even if calibrated to an absolute exposure metric. In fact, all of the exposure indices represent some statistic (e.g., mean or median) in some region identified on the IP. Therefore, it is important to recognize the exposure indices as only an *estimate* of the incident exposure on the detector, not an absolute value. In addition, exposure indices vary with kVp (peak kilovoltage) and beam filtration resulting from differential attenuation/absorption by the IP. Familiarity with the specific methods used by a given manufacturer are necessary for understanding the exposure index values, associating those values with an equivalent "speed" of the detector, and determining appropriate calibration methods.

4.2.1 Fuji Systems

Fuji PSP systems use a ***sensitivity number***[29] to provide an estimate of the incident exposure on the IP transmitted through the object. Under normal processing conditions for the standard resolution (ST) plates, the system sensitivity number for *nonfiltered 80 kVp beam* is calibrated to give:

$$S \cong \frac{200}{\text{exposure (mR)}}$$

A low (high) incident exposure with a low (high) PSL signal (determined from histogram analysis) requires increased (decreased) signal amplification to obtain an optimal signal range for digitization. The amount of amplification is indirectly indicated by the system sensitivity value, based on the exposure histogram shape. Computer algorithms identify the anatomically specific histogram shape; determine the minimum, maximum, and median values; and then map the median value to the center of the output range by adjusting the amplification as noted above. Two algorithms on the Fuji system include the *automatic* mode, which uses the whole area of the image for segmentation and histogram evaluation, and the *semiautomatic* mode, which evaluates a specific region within the image (e.g., a 10×10 cm^2 central area). The automatic mode also determines the latitude of the exposure based upon the range of the minimum to maximum histogram points, whereas the semiautomatic mode uses a predetermined latitude range. A third selection is the *fixed* sensitivity mode, where the user sets the sensitivity of the reader when

reading the IP. In this mode, the system responds similarly to a screen-film detector, and requires appropriate selection of radiographic technique.

4.2.2 Kodak Systems

Kodak PSP systems use an *exposure index (EI),*[19,47] a value directly proportional to the average log incident exposure on the plate, and is calculated as:

$$EI \cong 1000 \times \log(\text{exposure in MR}) + 2000$$

An exposure of 1 mR (80 kVp, 0.5 mm Cu, 1 mm Al filtration) results in an EI of 2000. An exposure of 10 mR leads to an EI of 3000, and an exposure of 0.1 mR will result in a value of 1000 for a calibrated system. Doubling the screen exposure results in an increase of 300 in the EI value; therefore, the units of EI are "kilobels" (analogous to decibels, commonly used in engineering). When using a high-resolution PSP detector ("HR" imaging plates), the EI has a lower range, resulting from less IP attenuation.

4.2.3 Agfa Systems

Agfa PSP systems utilize an exposure indicator called *"lgM,"* which is the logarithm of the median exposure value of the raw histogram.[48,49] Every Agfa PSP examination is assigned a Speed Class, and the system compensates for exposure variations of a factor of 4 around the intended speed. The lgM value indicates the actual exposure to the IP by a mathematical relationship to the *Scanned Average Level (SAL),* which is just the average grayscale value. A 2.2 mR (20 μGy) exposure to the IP using 75 kVp and 1.5 mm added Cu developed with a Speed Class of 200 results in an SAL of 1800. As a result of square-root amplification of the PMT output, the characteristic response of SAL with Speed Class of 200 is as follows:

$$SAL200 = 1214 \times [\text{exposure (mR)}].$$

The SAL value increases with the square root of the Speed Class, S, such that

$$SAL(S) = SAL200 \times (S/200)^{0.5}.$$

Grayscale values are remapped to a logarithmic scale of exposure, where 4095 equals 3.2768 and zero is undefined. The relationship between lgM and SAL is given by

$$lgM = 3.2768 - \log[(4095/SAL)^2].$$

Combining these three equations and simplifying, an exact relationship between lgM and exposure is revealed:

$$\text{exposure (mR)} = [(2276/S) \times (10^{(lgM - 3.2768)})].$$

A change of 0.3 in the numerical value of lgM corresponds to a change in exposure by a factor of 2. Although the absolute numerical value of lgM depends on Speed Class, the *relative* change between lgM and exposure is valid regardless of Speed Class. Therefore, lgM is expressed in "lgE" units, which correspond to *bels (B).*

This relative exposure paradigm is incorporated into dose monitoring software that is an option on Agfa PSP systems. For each examination, view, and cassette size, an average value of lgM is either calculated over 50 exams or established manually. For each subsequent examination of that type, view, and cassette size, the lgM value is compared to the nominal lgM. The dose offset is reported both numerically and in the form of a thermometer graphic. The nominal lgM values and average statistics for the last 100 exams of each type can be printed or made available in electronic form.

4.2.4 Konica Systems

Konica REX values[48] are generated from an exposure paradigm defined as follows:

$$S = QR \times E_1/E,$$

where QR is the preset quantization range, E_1 is the plate exposure in mR that produces a digital value of 1535, and E is the average exposure in the region of the plate used for the calculation of S. For a QR of 200, the system is calibrated to provide the 1535 code value with an exposure of 1 mR at 80 kVp to the IP. Thus, a 1 mR exposure yields an S of 200 × (1/1) = 200, and 2 mR yields an S of 100.

The manufacturers are universally basing their "target" exposures on an incident exposure of 1 mR to the IP (even though energy dependence varies), which is close to a "200-speed-equivalent" class detector. This is likely due to the recognition of less absorption efficiency of the PSP IP relative to a 400-speed screen-film detector (by about one-half) and the need to achieve a similar SNR on the image relative to the screen-film image. Another point is the difference in the beam quality used for calibration by the manufacturers. There is a strong dependence on kVp and filtration for consistency of the exposure indices for all systems. An effort to standardize the methods for exposure index calibration for all DR devices is needed and is currently being undertaken by the AAPM.[50]

4.3 Exposure Concerns When Using PSP Systems

The exposure indicator estimate of the incident exposure to the PSP detector is sensitive to segmentation algorithms, anatomical menu or histogram selected, amount of collimation (or lack thereof), effective energy of the beam (kVp, filtration), positioning of the patient relative to the phosphor, the presence of high-density objects in the field of view (e.g., prosthetic implants), the source-to-image distance, and the delay between exposure and readout, among other factors. Because the PSP system provides a nearly optimal display of the anatomical information *independent of exposure,* this number is an important aspect of quality assurance, patient care, and training issues. The optimal exposure range for clinical imaging procedures such as chest imaging requires an x-ray technique corresponding to a ~200-speed screen-film detector system, based upon the empirical analysis of images and characteristics of the PSP image acquisition process.[27,51] For extremities, a higher exposure should be considered (e.g., 50 to 100 speed, similar

to extremity screen-film cassettes), while for pediatric imaging a lower exposure is recommended (e.g., 400 speed or faster, depending on the radiologist's concern for dose and tolerance of quantum mottle in the image). Underexposed PSP images are identified by increased quantum mottle due to insufficient x-ray photons and subsequent amplification that reduces SNR and contrast resolution. In selected examinations, radiation exposures can be reduced when the signal is sufficient in the presence of increased noise (e.g., nasogastric tube placement,[27] scoliosis follow-up examinations[43]).

More problematic are overexposures, due to the large dynamic range of the "overexposure" region (see Figure 11), the inability to easily detect the overexposure (the image looks great), and the potential for complacency on the part of both the technologist and the radiologist in accepting these images without understanding the disservice to good patient care and proper radiation safety. Visual cues and information on each printed film or soft-copy image should be available to alert radiologists and technologists that the exposures are outside "normal" limits. Optimal radiographic techniques for PSP detectors might differ from screen-film detectors, particularly for the kVp setting because of the inherent differences in the phosphor composition and digital post-processing of the image.[52] In general, the recommended kVp setting for PSP detectors is slightly higher than screen-film for imaging small thicknesses (e.g., extremities, young pediatric patients). For instance, instead of 60 to 65 kVp, the use of 5 to 10 kVp higher (65 to 75 kVp) will lower patient dose (only with a commensurate decrease in the mAs for the same incident x-ray fluence on the detector) but will not significantly affect the *processed* image contrast. Similarly, for imaging large thicknesses in which the screen-film technique is typically 100 kVp or higher (e.g., adult chest), a suggested corresponding technique for PSP detectors is 5 to 15 kVp less. This is chiefly due to the poorer absorption of the phosphor compared to the screen-film phosphor at higher effective energies. A suggested kVp for an adult chest image is 100 to 110 kVp, compared to 120 to 130 kVp for a corresponding screen-film image. These are general recommendations *only,* and specific techniques must be determined on a case-by-case basis, preferably with the feedback from a radiologist.

Technologists should be advised to adjust manual radiographic techniques for grid and no-grid examinations by taking account of the grid Bucky factor. It is very easy to become complacent and use a grid technique when the grid is not used, particularly for portable exams, producing a needless overexposure by a factor of 2 to 3. Continuous training, in-services, oversight, and feedback are recommended to ensure proper use and radiographic technique with PSP detectors.

5. PSP SYSTEM IMAGE CHARACTERISTICS

5.1 Spatial Resolution

Limiting high contrast resolution in PSP radiography is determined by several factors: physical limits imposed by the composition and thickness of the phosphor plate, the size of the laser spot, temporal lag of the PSL, and light scattering within the phosphor contribute to the modulation and loss of the "pre-sampled" signal. The finite diameter of the laser spot and the spread of PSL, particularly at depth, contribute to unsharpness, as shown in Figure 18. Pixel size, typically between 100 and 200 μm, determines the maximum spatial resolution of the system, up to physical limits imposed by the composition of the IP and the size of the laser spot. Digital sampling confines the accurately depicted spatial frequencies to a maximum called the Nyquist frequency,
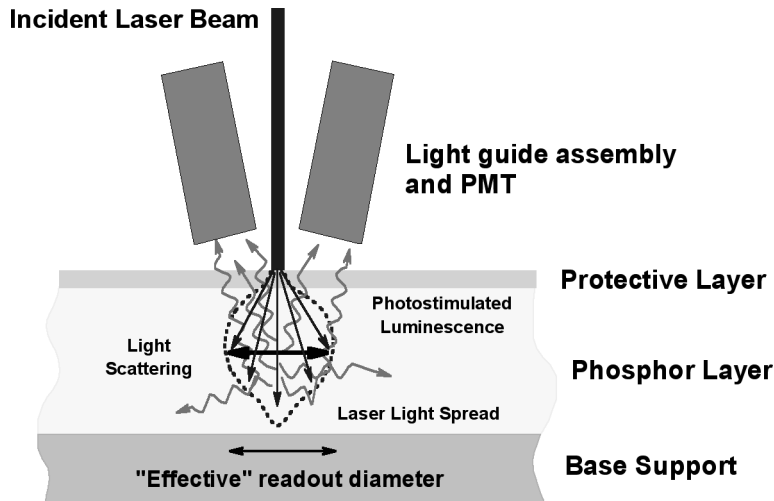
Figure 18. The effective area of the phosphor simultaneously stimulated by the laser is determined by the incident laser diameter, laser light spread within the phosphor, and the distribution of the PSL collected by the light guide assembly. This spread reduces the modulation of higher frequency signals. (Adapted from reference 17, page 746, fig. 15.)

equal to the inverse of twice the pixel length, $(2\Delta x)^{-1}$. Unlike conventional screen-film cassettes, smaller phosphor plates will sometimes provide better limiting resolution than larger plates when the number of samples is constant, independent of the IP dimension (more often in newer systems, the sampling pitch is constant, and any changes in output resolution or matrix size is dependent on system configuration settings). Resolution sharpness can be increased with the use of a thinner phosphor layer using high-resolution PSP detectors (see Figure 19); however, lower detection efficiency requires a higher radiation dose. Phosphorescence lag causes the spatial resolution in the fast-scan direction to be slightly less than that in the sub-scan direction.
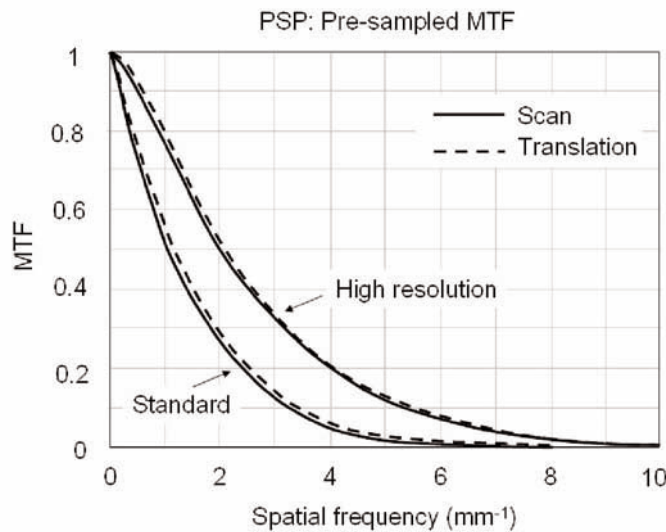
Figure 19. Typical results for pre-sampled MTF curves with PSP detectors are illustrated. The curves on the left are for standard resolution (thick phosphor) and on the right for high resolution (thin phosphor). Solid and dashed lines distinguish the scan and sub-scan MTFs, respectively. (Adapted from reference 52).

Aliasing represents high-frequency signals contained in the image above the Nyquist frequency, f_N, reflected back at lower spatial frequencies. A sampling aperture of 0.2 mm gives $f_N = 2.5$ mm^{-1}, and a high-frequency signal such as a stationary grid pattern at 3.6 mm^{-1} is 1.1 mm^{-1} above f_N. The grid pattern will be reproduced at frequencies 1.1 mm^{-1} below f_N, at 1.4 mm^{-1}. The impact of aliasing increases image noise and reduces the detective quantum efficiency (DQE) of the PSP detector. Notable examples are the aliased signals generated by stationary anti-scatter grids with lead strip frequencies beyond the Nyquist frequency, and improper sub-sampling of the image, as illustrated in Figure 20. Aliasing can be reduced in the (fast) scan direction with a mathematical filter to reduce or eliminate these high-frequency (aliased) signals. It is not possible to implement a low pass anti-aliasing filter in the slow scan direction as the lines are acquired one at a time. Aliasing is most likely to occur when the grid lines are oriented parallel to the fast scan direction.

5.2 Contrast Resolution

The minimum difference in a "noiseless" signal that can be represented between digital pixels in the image depends on the total number of possible code values (quantization levels), as well as the target signal amplitude relative to the background. In most PSP systems, pixel values change with the logarithm of PSL, or equally with the logarithm of radiation dose to the plate, so the numerical difference between pixel values is the contrast. *Contrast* resolution of a PSP system depends not only on the number of bits used to represent each pixel, but also by the gain of the system (e.g., number of electrons/x-ray photon, number of x-ray photons per analog-to-digital code value) and overall noise amplitude relative to the contrast difference. The ability to detect a specific signal in the image is strongly dependent on the inherent subject contrast (kVp, scatter acceptance), amount of noise (x-ray, luminance, electronic, and background variations), image viewing conditions, applied image processing, and the limitations of the observer in discerning distinguishing regions of low contrast and small object size.

PSP contrast resolution is similar to the screen-film image, but is limited by digital image noise as opposed to the speed and latitude of the film. Separation of latent image and display processing stages allow application of examination-specific contrast enhancement, which otherwise is extremely low (see the characteristic curve response in Figure 11). Unlike screen-film detectors, which are contrast *limited* at a particular radiographic speed (the classic trade-off

Figure 20. A stationary antiscatter grid can cause aliasing, where the high-frequency patterns are reflected back into the low-frequency spectrum about the Nyquist frequency as superimposed beat patterns.

between detector latitude and film contrast), the PSP image contrast is *noise limited.* The random variation of absorbed x-rays in the PSP detector determines the quantum noise component. *Stimulated luminance variations* during the readout process contribute significant variations in the output signal. *Electronic noise* sources cause a further variation in the output signal. *Quantization noise* adds inaccuracies in the determination of the discrete digital signal amplitude values, dependent upon the bit depth of the ADC, typically 10 to 12 bits in current systems. To approximate the typical image noise in a 400-speed film (and thus achieve similar contrast sensitivity), the PSP detector (using standard resolution plates) requires a higher x-ray photon flux by about a factor of 2 times (e.g., a 200-speed system).[27] Lower detection efficiency of the phosphor plate relative to a typical rare-earth dual-screen cassette is the chief cause. In addition, edge enhancement processing may influence the appearance of noise.

5.3 Detective Quantum Efficiency (DQE)

The DQE describes the efficiency of information transfer from the transmitted x-ray fluence on the detector to the output image used for diagnosis with respect to spatial frequency. It is dependent on the quantum detection efficiency (QDE) of the image detector and the conversion efficiency (CE) and associated noise with each stage involved in creating the final image. This includes the number of trapped electrons per absorbed x-ray photon, noise in the stimulation and emission of the latent image, noise in the conversion to an electronic signal, noise associated with the digitization, and noise occurring in the final output image presentation. An estimate of the large area, zero frequency DQE of a storage phosphor has been formulated as:[53,54]

$$DQE_{PSP} = \frac{X_{abs}}{[1+CV(E)][1+CV(el)][1+CV(S)]+<g>^{-1}}$$

where: X_{abs} = fraction of incident x-ray photons absorbed in the phosphor layer,

$CV(E)$ = coefficient of variation of the x-ray energy absorbed in the phosphor layer,

$CV(el)$ = coefficient of variation in the number of trapped electrons for a given absorbed energy,

$CV(S)$ = coefficient of variation of the light signal emerging from the phosphor for a given number of trapped electrons, and

$<g>$ = the average number of photoelectrons detected at the photomultiplier per absorbed x-ray (the large-area response function).

The energy dependence of X_{abs} is plotted in Figure 4. *CV(E)* depends on the overlap of the spectrum with the k-edge of barium and the fraction of K characteristic x-ray escape. For an 80 kVp x-ray beam transmitted through the patient, a value of ~0.15 has been estimated for the coefficient of variation of energy absorption in the phosphor layer. Hundreds of electrons are trapped in the phosphor F-centers per absorbed x-ray photon, making *CV(el)* relatively small (<0.05). On the other hand, the variation of stimulating laser light with depth in the phosphor and a similar variation of emission light with depth makes the luminance noise value *CV(S)* quite high, estimated to be on the order of 0.8.[23] Large-area gain in the phosphor, *<g>*, is ~10, and results in the value in the denominator of the DQE expression ≅ 2. Thus, the DQE(0) can be estimated as approximately 1/2 X_{abs}. At 80 kVp with a typical transmitted x-ray spectrum through a patient, DQE(0) ≅ 0.25 for standard resolution and DQE(0) ≅ 0.13 for high-resolution phosphor

plates. These values approximate the published findings of Dobbins[52] and Hillen.[55] Values of DQE(f) has been thoroughly investigated for several generations of storage phosphor imaging plates,[52,56] showing a slow, but steady improvement in the development of phosphor plate technology and subsequent detection efficiency as a function of spatial frequency. Recent introductions of dual-side readout[13,18] and structured phosphor technologies[14,22] for PSP detectors have demonstrated great improvement in the DQE, approaching that of flat-panel detectors.[20,21]

In a digital system with sufficient bit depth and resolution, increased SNR can be obtained by simply increasing the radiation exposure to the detector up to a saturation level or to a point where other noise sources begin to dominate, but the cost is increased exposure to the patient. Quantitative detector analysis can determine system performance using objective measurements. The *pre-sampled* modulation transfer function, MTF(f), is a measure of the detector object transfer efficiency as a function of spatial frequency (the *signal*). The noise power spectrum, NPS(f), is a measure of the noise characteristics of the detector (the *noise*). When scaled by a conversion constant, a ratio of $MTF^2(f)$ to NPS(f) generates the noise equivalent quanta, NEQ(f), as a function of spatial frequency. This is an estimate of the *equivalent* number of x-ray photons per unit area (usually mm^2) that the detector effectively uses. With increasing incident exposure, the NEQ(f) similarly increases to a point where the system saturates and/or other noise sources dominate, after which NEQ(f) decreases. At a given spatial frequency, the measured NEQ(f) represents the output signal to noise ratio, SNR^2_{out}.

The DQE is the ratio of the noise equivalent quanta to the actual number of quanta as a function of spatial frequency: DQE(f) = NEQ(f) / q. The incident radiation flux, q, is the number of x-ray photons incident on the detector per unit area (usually expressed in mm^2) determined from computer simulations from a known x-ray source, kVp, mAs, and beam quality (half-value layer [HVL]).[57] DQE is also calculated as: SNR^2_{out} / SNR^2_{in} . A perfect detector system has a DQE(f) of 100% at all spatial frequencies. X-ray detector systems lose efficiency over smaller areas (at high spatial frequency) because of the inability of the detector to efficiently capture x-ray information and/or have additive noise such as electronic amplification noise, pixel drop-out, or phosphor "structure" noise (among other sources), all of which can mask the true signals. Some digital detectors can use the incident radiation more effectively, and thus can reduce the exposure to the patient for a given SNR. In general, DQE(f) measurements for digital detectors range from less than 10% to as high as 80% for large-area objects (low spatial frequency range). The actual DQE depends on the detector design and x-ray converter characteristics. For high spatial frequencies (detail information) the DQE drops to the point where the system can no longer retain the identity of a small-area input signal. To be useful, it is important to determine the incident exposure range over which the DQE is quoted because some systems operate more effectively over a given exposure range (e.g., mammography requires much more radiation exposure than conventional imaging). Knowledge of the DQE, NEQ, NPS, and MTF for PSP radiography allows objective comparisons that can assist in the determination of appropriate and reasonable performance for a particular imaging application. With that being said, measurement of the NPS and MTF is beyond the scope of the clinical acceptance testing of a PSP system.

5.4 Image Display

Manufacturers design image processing parameters that assume the shape of the LUT to be applied by the display device. In a networked environment, it is important to confirm that the

display device, be it a printer, or a PACS workstation, is applying the same LUT as the PSP system vendor expects.

Cathode ray tube (CRT) **and** *flat-panel* **monitors** are used for "soft-copy" display. The accuracy of grayscale rendition is crucial, from the QC workstation of the technologist to the interpretation monitor for the radiologist or referring physician, to ensure the optimal transfer of image information. Monitors are often the weak link in the overall digital acquisition and display system. Specific calibration procedures have now become standardized, as documented in DICOM PS3.14, the Grayscale Standard Display Function (GSDF).[58] This standard is based upon the perceptually linear response of the human observer, and the ability to change the digital driving levels of a monitor to ensure that the grayscale that is rendered follows that response. Acceptance testing and quality control are specified in the AAPM Task Group 18 document on display monitor specifications, measurements, and quality control.[59] An increased emphasis on the acceptance testing and quality control of display devices and viewing conditions is necessary to ensure optimal image rendition, as any x-ray imaging *system* is only as good as its weakest link. The report of AAPM Task Group 18 describes functional characteristics, acceptance testing, and quality control of display technologies for medical imaging applications. It is extremely important to ensure the proper operation of the display devices as part of the overall quality assurance and quality control program.

Laser film printers convert the digital images to film images to mimic the conventional screen-film paradigm, where the film is transilluminated for viewing. With some PSP systems, the image size must be reduced (demagnified) by a variable amount, depending upon the phosphor plate size and output film format. Hard-copy presentation of the PSP image commits the user to a single rendering, obviating a major advantage of digital display processing. In order to provide two different grayscale/edge enhancement renderings, the image size may be further reduced to accommodate two images on a single film. This two-on-one format requires a reduction to 50% of the 35 × 43 cm (14 × 17 in.) FOV on small-format PSP films (~26 × 36 cm). Size reductions complicate direct measurements on film and make comparisons of films with size differences more difficult. Full field of view printing is available on 35 × 43 cm format film, with sampling matrices up to approximately 7000 × 8600 pixels (by one manufacturer) to provide high spatial resolution typically of 5 to 10 line pairs per millimeter (lp/mm) over the full FOV. In networked laser printers, large film printing is available for a range of digital matrix sizes by interpolation and extrapolation of the digital image data. Slight size reductions of 5% may occur with many laser printers in this large format to accommodate a border around the film, an important detail to know when size measurements are made from the film, for example in orthopedics.

Printer calibration requires matching of the grayscale appearance of the image on the display with that on the film. Conventional methods involve working with the printer manufacturer to use predesigned LUTs specific to a particular PSP device to produce similar looks. An alternate approach is to use DICOM presentation values (p-values), on a printer calibrated to the DICOM standard, that adhere to the standard and that through the use of appropriate DICOM-compliant LUTs can be matched to display devices operating at much lower maximum luminance levels. In addition to this, CRT phosphors and liquid crystal display (LCD) backlights produce different colors and CRTs have different phosphorescence lag when changing images.

The adverse effect of high ambient light levels on the appearance of the image is more problematic with a soft-copy display device than with a transilluminated film because of the lower luminance of these devices. Steps to control ambient lighting in the QC console area must be taken. While this situation is complex, DICOM PS3.14 provides methods to address the differences in luminance levels.

6. GENERIC FUNCTIONAL SPECIFICATIONS OF PSP SYSTEMS

It is highly recommended to communicate with marketing specialists and system engineers to determine the up-to-date capabilities/specifications of a particular PSP system prior to purchase, installation, and testing. A specific system configuration can significantly affect how the physicist conducts acceptance tests. Functional specifications related to "typical" capabilities/specifications are listed based upon a review of vendor literature.

6.1 Phosphor Detectors and Cassettes

Several detector and cassette sizes are available for PSP systems. The most popular sizes include 35 × 43 cm (14 × 17 in.), 35 × 35 cm (14 × 14 in.), 24 × 30 cm (10 × 12 in.), 24 × 24 cm (10 × 10 in.) and 18 × 24 cm (8 × 10 in.). Special size cassettes (e.g., 20 × 20 cm) and those needed for specialized applications (e.g., scoliosis, long FOV, and dental panorex x-ray) are also available as options from certain manufacturers. The time required reading the phosphor plate is dependent on the plate size. Larger sizes usually take longer to read, and decrease overall system throughput. Spatial resolution can also be affected by phosphor plate size. The dependence of spatial resolution is highly manufacturer-dependent, and is sometimes related to the IP dimensions. Configuration of output spatial sampling and matrix size may also differ from the intrinsic resolution, with larger resolution elements and smaller matrices often used for large FOV detectors, providing adequate resolution and detail for the particular studies. Plate inventory should be sufficient to eliminate delays due to accessibility of plates, and not by the throughput of the PSP reader. Related to the second item, it is highly recommended to have two or more PSP systems in busy work environments for redundancy in the event of system malfunctions or scheduled downtime. Standard resolution and high-resolution image detectors are available from the manufacturers, and should be considered for use relative to the imaging application. A trade-off of detection efficiency for slightly better spatial resolution requires approximately three times more exposure with the high-resolution detector to achieve an equivalent SNR.[19] (See section 6.3.)

6.2 PSP Detector Throughput

Throughput is defined as the average time from the insertion of an imaging plate/cassette to the time it can be reused. A range of ~30 up to ~200 IPs per hour are specified by the various manufacturers for laser point scan systems, depending on the equipment and options purchased. PSL decay time is the major limit to the throughput speed, although in some systems plate handling and erasure requirements can add substantial time for the transit of a plate through the reader. Some PSP systems have internal stackers or external automatic handling capabilities to allow the user to accomplish other functions without having to wait for the total readout process. Multiple plate readers take advantage of "pipeline-processing" and simultaneous reading and erasing to provide an average higher throughput than "one-plate" readers. Line-scan systems provide even higher throughput but are typically part of a detector device slaved to a specific purpose, e.g., chest stand or under-table Bucky device. In terms of enhanced efficiency, line-scan systems from the various manufacturers are expected to increase in use, replacing systems that would otherwise use a cassetteless point-scan system.[22,23,28]

6.3 Spatial Resolution

Spatial resolution is chiefly dependent on the reading and recording laser sampling pitch over a given FOV (phosphor plate size), which determines pixel size. Many PSP reader systems utilize a laser beam with an effective 100-μm diameter spot size on the phosphor, but there are variations from this value. The output sampling pitch is determined by the element length and the number of elements across the detector. In some cases, the element length is larger for larger detectors with the number of elements remaining the same. PSP detector characteristics including phosphor coating thickness, protective coating layer thickness, finite laser beam dimensions, and light scatter in the phosphor, as well as frequency response of electrical circuits will determine limiting spatial resolution potentially greater than that expected from resolution element size. In general, limiting spatial resolution (<5% MTF) ranges between 2.5 to 5 lp/mm (0.2 mm to 0.1 mm object detail), which is less than a 400-speed screen-film resolution capability of about 4 lp/mm (0.125 mm object detail) at 20% MTF.[60] Resolution of 10 lp/mm, equivalent to 0.05-mm detector element length is available with some PSP systems that are designed for mammography and high-resolution radiography.

There are several generations of PSP detectors with different physical and performance characteristics available from the manufacturers. "Standard-resolution" and "high-resolution" plates that can be used in the same PSP reader are provided by some manufacturers. The former is typically used for all applications in general radiography, while the latter is used for extremities and proposed mammography applications. The thickness of the standard-resolution plates is approximately two times greater than that of the high-resolution plates. As with screen-film detectors, there is a trade-off of QDE and contrast resolution for spatial resolution for a specific exposure. To achieve standard- or high-resolution output, the sampling pitch and laser spot size are often changed. Information of resolution factors, MTF measurements, and DQE measurements are in the literature.[17,52]

Soft-copy display devices also influence spatial resolution of the displayed radiograph. CRT and LCD monitors can be the limiting factor for displaying image matrices, depending on the FOV, the bandwidth and number of television lines of the monitor, or the number of pixel elements. Images with a larger matrix size than the display can support require pixel averaging with a corresponding loss of detail. One-to-one image pixel to display pixel mapping is necessary for achieving true intrinsic spatial resolution on the output display, which often requires a portion of the image to be displayed at one time, thus sacrificing the FOV of the whole image. With image panning, the whole image can be investigated at the intrinsic resolution limit of the detector.

6.4 Contrast Resolution

Optimally tuned PSP systems with low electronic noise will have contrast resolution chiefly determined by image acquisition techniques (kVp, antiscatter grid, geometry, etc.), processing parameters (display gradient, frequency processing, noise filtering), and DQE of the PSP phosphor. Image information thus acquired without post-processing is termed "raw data" and represents grayscale values that correspond to the PSL emitted from the screen. These images are totally unacceptable to the human viewer, due to the lack of display contrast that is achieved by data ranging and post-processing, two crucial steps for optimizing contrast resolution.

A major equipment issue is the bit depth of the pixel, which determines the number of discrete gray levels that encode the contrast differences. Ten bits has been shown to be sufficient

for film recording,[17] and most soft-copy monitors (either CRT or LCD) use a signal derived from an 8-bit range of values. In most PSP acquisition systems, 12 or more bits are used for the *initial* digitization, the number of which is vendor specific. Regardless of the acquisition bit depth, there is a potential to lose image information if scaling algorithms and/or histogram analysis are improperly applied, or if the display medium (hard or soft copy) has inadequate or noncalibrated gray-level display capabilities. Contrast resolution is dependent on the incident exposure; the dynamic range; the signal derived from attenuation differences; and the noise defined by the quantum, electronic, quantization, and detector variable response uncertainties. The signal is dependent on the beam quality and effective attenuation coefficient of the objects in the beam path. The noise should be chiefly dependent on x-ray quantum statistics over the clinically relevant exposure range, where the greatest contribution to the statistical uncertainty is the x-rays contributing to image formation. This requires that the PSP detector/reader be optimized to have other noise sources significantly smaller that quantum noise. A minimally acceptable SNR level allows contrast enhancement through image post-processing without excessive amplification of the noise.

In terms of image output, two alternatives for display of processed images include "for presentation" and "for processing." For presentation images, use a "burned-in" LUT applied by the modality workstation that defines the useful exposure range determined by the "autoranging" function and contrast enhancement over the region. Information below or above the predefined range is lost due to thresholding or saturation. "For processing" images have the full dynamic range information content maintained, and a nonlinear (e.g., sigmoid-shaped curve) value-of-interest look-up-table (VOI LUT) provides the conversion of raw data to display of contrast enhanced data on-the-fly,[61] and allows the user to adjust the display window outside of the boundaries set by autoranging algorithms. In a PACS environment this allows one to overcome mistakes in the initial analysis of the histogram data or by user input errors in selecting the correct examination algorithms. A PACS-aware "for processing" image capability with VOI LUT is necessary to use this capability; otherwise the images will appear very flat and washed-out. Without VOI LUT or vendor-specific enhancement, "for presentation" data should be sent to the PACS.

6.5 Dynamic Range

The incident exposure sensitivity of the PSP detector typically extends from 0.01 mR up to 100 mR (a range of about 10,000 or 10^4, determined by the amplifier settings of the PMT). In some systems, a "high gain" setting can reduce the lowest detectable exposure to 0.001 mR. Logarithmic amplification is applied in many systems, partially to compress the dynamic range of the exposure-luminance response curve, and to use the limited output integer range more effectively. (Note: In some systems, "square-root" amplification is used in lieu of log amplification. See section 2.3.6, Digitization.) Intrinsic detector contrast is low, and is not clinically optimal for a human viewer. ("Four decades" of dynamic range is attributed because of this tremendous exposure response; however, rarely are four decades of dynamic range required *or desired* for diagnostic radiology applications.) The range of exposures containing the *useful image* information is identified with image analysis of the digital distribution on the raw image, usually by *histogram analysis*. Examination-specific algorithms evaluate the distribution and shape of the resultant histogram, followed by contrast enhancement to mimic screen-film response in "for presentation" images tuned for radiologist preference. However, for images submitted to a computer viewer, preferred is the unprocessed or "for processing" output.

6.6 Desirable Specifications and Features

6.6.1 Phosphor Plates, Cassettes, Grids, Identification Terminals

The number of phosphor plates and cassettes should meet 1.5 times the peak simultaneous demand for imaging services, for all IP sizes. Usually, only a minimal number of IPs is provided in the basic CR reader package. Stationary, low-frequency grids can be problematic with digitally sampled images, including PSP systems. High-frequency grids of 55 lines/cm (140 lines/in.) up to 70 lines/cm (178 lines/inch) and multi-hole grids are available to alleviate problems with aliasing and moiré patterns, particularly in the slow-scan direction, and should be considered as part of the system purchase. Cassette identification (ID) terminals electronically correlate the patient to the cassette, and provide the examination-specific processing instructions. ID terminals placed in convenient, strategic locations in the work environment are important to avoid workflow bottlenecks and throughput problems. In addition, ID terminals and PSP readers within a specific locale should be locally networked to allow any identified IP to be processed by any reader for redundancy and increased throughput.

6.6.2 Output Hard-Copy Image Characteristics

Output image format of 1:1 magnification for all image sizes (18 × 24 cm through 35 × 43 cm) should be requested. Many laser printers reduce the actual image size by up to 5% (95% of "true" size that would be achieved on a screen-film detector). Distance calibration marks on the side of the films should also be included. Laser printers calibrated for DICOM GSDF presentation state are preferred, and PSP image review workstations that send presentation values ("p-values") to the printer will avoid the modality-specific LUTs that are part of a vendor's laser printer setup.[61] If GSDF calibration is specifically not available, it is prudent to request user-adjustable printer LUTs to match grayscale appearance on the soft-copy monitor.

6.6.3 Incident Exposure Estimation; Other Data Fields

Incident exposure estimates for PSP image acquisitions are extremely important, should be included on the image demographics as a requirement, and preferably tracked in a database for long-term analysis of trends. In addition, other performance indices and database functions should be considered, including display of extremely high or low exposures, number of exposure cycles of each IP (to track longevity), and processing parameters applied to the image, among other data fields.

6.6.4 Image Processing Functionality

Specific image processing capabilities should be listed. These include simple window/level adjustments, nonlinear adjustments to mimic screen-film response, reverse contrast mapping, edge enhancement, dynamic range control, and fill-in "surround" of unexposed areas (e.g., to fill in the unexposed areas of the resultant image with dark or opaque boundaries—crucial for pediatric newborn studies and small objects). The ability to implement user-defined functions in addition to the built-in functions is desirable. All image processing should be compatible with displays and printers that conform to DICOM PS3.14 and use the GSDF calibration and presentation state.

6.6.5 Patient Demographics and Film Marker Positioning

Systems should have the flexibility of allowing specific institutional, patient, phosphor plate, and technologist identification with a user-selectable font type/size and position on the image. Image processing parameters, exposure index, image magnification or minification, image reversal, and positioning markers should be available.

6.6.6 PSP System Interfaces to RIS, HIS, and PACS

In a PACS environment, insist upon DICOM Modality Work List (MWL) for the ID terminal or user workstation at the acquisition device, which is most often provided by a PACS "broker" (interacting via HL7 with the Radiology Information System/Hospital Information System [RIS/HIS] and DICOM with the PACS). A list of scheduled patients and patient demographic data allows easy and efficient identification of the exposed IP/cassette (keystroke entries are often the cause of broken studies on the PACS) by tagging the patient and exam information in the DICOM image header. Room and patient scheduling can be achieved through "scheduled workflow profiles" described in the Integrating the Healthcare Enterprise (IHE) guidelines.[62–68] Newer DICOM elements such as performed procedure step, VOI LUTs, and presentation state capabilities[1] are components that the PSP *and* PACS vendors should both implement (at the time of publication these capabilities are not universal). A DICOM conformance statement for the PSP reader, PACS broker, QC workstation, laser printer, and other peripherals is necessary to verify desired functionality and information transfer from the PSP system to the PACS and other pertinent peripherals.[69] Proprietary data delivered in "private tags" (odd numbered elements) and desirable optional information in the DICOM header should be specified in the negotiations for the equipment as these items are often inaccessible otherwise. Knowledgeable PACS and informatics expert assistance is helpful for such complicated issues. There are also capable third-party vendors providing functional interfaces between PACS and PSP systems.

6.6.7 Quality Control Phantom; QC Workstation and Software

The vendor/manufacturer should provide a quality control phantom and evaluation program with the PSP system or systems. This is often an option at additional cost for the hardware and software, but it is necessary for the user, and is highly recommended. Spatial resolution, contrast resolution, exposure uniformity, exposure linearity, and distance measurement accuracy are the tests that can provide trend analysis and reliably demonstrate compliance and failure. Ideally, a third-party phantom, specifically designed for "generic" PSP acquisition performance and image quality should be specified in addition to the manufacturer-specific phantom at larger, multivendor sites. The QC/image review workstation allows verification of patient positioning, image orientation, and proper grayscale rendition. Monitor calibration is crucial to maintain consistent appearance at the PSP review workstation as well as the PACS and other soft-copy display workstations. This can be achieved by implementing DICOM PS.14 GSDF.[58] Luminance (≥ 171 cd/m^2 [candela per square meters]) and luminance ratios ($\geq 100:1$) as well as low room illuminance <20 lux (sometimes difficult to achieve) should meet the recommendations of AAPM TG-18.[70] Ideally, the technologist workstation should have capabilities meeting the minimal standards of a primary diagnostic review station with luminance ≥ 250 cd/m^2 and luminance ratio $\geq 250:1$, particularly for systems in which the image contrast is adjusted by the user. Image monitors, particularly LCD, should be carefully inspected to ensure minimal angular dependence, particularly in the vertical direction (from top to bottom), as image appearance can drastically change with

viewing angle, which causes unintended variations in window/level settings by technologists of various heights. The QC workstation should have software tools to assist in measuring system performance. Manual and automated quantitative tools such as region of interest (ROI) pixel values, ROI standard deviations, contrast-to-noise ratios and maximum SNRs for QC image tests are desirable. Periodic results cataloged in a database and plotted versus time (daily, weekly, and monthly) provides trend analysis and performance indices to show compliance failures, and can indicate the need for preventive maintenance prior to a problem being manifested.

6.6.8 Service Contracts, Preventive Maintenance, Warranty, and Siting Requirements

Hardware/software upgrades, phosphor detector/cassette longevity, and system extended warranty considerations should be part of the maintenance contract. This includes approved third-party service or in-house radiological engineering support/training as primary responders if desired. Site preparation issues include details about required power, air conditioning/filtering, equipment footprint, configuration of the PSP readers, preliminary schematic drawings, etc. The warranty terms should specify the expected on-site response time, and time to guaranteed resolution of any reported maintenance issue. Penalty clauses for failure to meet expected levels of service and maintenance should be specified.

6.6.9 Application Training for Technologists, Radiologists, Physicists, Clinical Engineers

Specific reference to applications training should be indicated, even when the vendor has a standard level of applications training with the sale of the equipment. A minimum of 1-week, on-site training is recommended for technologists (include work shift hours as necessary). This should be followed by a subsequent week of refresher assistance approximately 1 to 2 months after the initial training. Often, a super-tech is specified for advanced training at the manufacturer's facility. Radiologists should interact with the application specialist during the initial startup of the system to implement specific image processing algorithms appropriate for each examination. Physicists should be aware of processing algorithm tuning functions and be instructed on processing variables, effects on image appearance, and adjustment procedures. Hospital engineering staff should be trained for simple preventive maintenance tasks and error recovery issues. In addition, these individuals should also have the option of attending a training program designed for preventive maintenance and in-depth system repairs, particularly in the absence of a warranty agreement.

7. CLINICAL IMPLEMENTATION ISSUES

7.1 Expectations and Realities

The benefits anticipated by the introduction of PSP systems into clinical practice depend on the intended role. When PSP radiography is introduced as a replacement for screen-film as a detector, the user expects to benefit from the improved consistency and decreased repeat rate when compared to conventional radiography. When PSP radiography is introduced to provide flexibility in the presentation of the image, the user expects to benefit from the ability to reprocess the digital image. When PSP radiography is introduced in order to replace film with a digital archive, the user expects to benefit from the convenience of storing images in electronic form.

When PSP radiography is introduced to replace film with a digital image distribution and display system, the user expects to rely on PSP images exclusively for acquisition of ordinary radiographic examinations.

The clinical acceptance of PSP radiography depends in part on perceptions of comparisons with conventional screen-film for similar tasks. A benefit of PSP radiography over screen-film is the ability to modify the appearance of the digital image to enhance conspicuity of clinical features. However, there is no universal agreement on the optimal set of processing values for an examination. The effects of display processing depend on radiographic technique, and too much "processing" can generate undesirable results. The ability for a knowledgeable operator to modify processing defaults introduces a configuration management problem: for consistency, it is important to assure that the same processing defaults are loaded in all PSP machines and identification terminals. Differences in processing methods between manufacturers mean that PSP images produced by identical radiographic technique with the same subject may not have the same appearance with PSP systems of different vendors. Comparison examinations over time should have the same image processing parameters to assist consistent clinical diagnosis.

7.2 Technical Concerns

Because the first step in processing the PSP image is to locate the exposure field and ignore signals outside the field, patient positioning, x-ray beam collimation, and convergence of the light field and x-ray field are more critical than in screen-film imaging. Generally, the anatomy of interest should be centered on the IP, collimation should be used to reduce the amount of beam that is unattenuated by the subject; collimation should be symmetric and parallel to the edges of the cassette. When an image is acquired without collimation borders on one side of the IP, the finding routine will often fail to recognize the correct borders and produce an unrealistic histogram, and cause scaling errors. Sometimes the anatomy of interest cannot always be centered on the IP and collimation cannot be symmetric or parallel, for example, elbows. Current technology (year 2000 and beyond) PSP systems are more tolerant for such cases. Additionally, it is prudent to acquire only one image per phosphor plate, but there are some systems that can effectively deal with multiple exposures. When one considers the potential for confusing the PSP reader with multiple images per plate, and a somewhat better spatial resolution for smaller cassettes, it is advised to project a single view on the smallest cassette possible, especially for extremity exams emphasizing bone rather than soft tissue. If digital images are to be viewed on a soft-copy device, separate images can be manipulated independently, unlike multiple images on a single detector (display). If multiple views are to be used, it is helpful to place only similar views on a given detector. The PSP reader algorithms may be able to segregate the views for histogram analysis, but appropriate display processing selected for one view must be applied to all.

Unlike screen-film radiography, the size of the cassette selected can have a pronounced influence on the image resolution, particularly for older PSP readers, where the output matrix is fixed and the spatial sampling is adjusted to fit the detector dimension. Newer systems with fixed spatial sampling have an image matrix size that varies according to the imaging plate dimension, and spatial resolution is independent of the detector size. In some cases, the output image sent to the PACS is reduced by pixel averaging the acquired image at the image QC workstation. For instance, a 35 × 43 cm image with 100 μm sampling (3500 × 4300) is often converted to 200 μm (1750 × 2150), which reduces resolution by a factor of 2 but reduces image size by a factor of 4. There is another effect for hard-copy films, depending on the format of the laser

printer and the size of the films: the images acquired with the smaller cassettes may be presented at 100% magnification, while those from the largest cassette may be presented in reduced size.

IPs of different generations differ in their x-ray capture and light-generating characteristics, thus the PSP reader should be calibrated specifically for a single detector type. Some detectors may not be appropriate for a particular PSP reader model because of laser light characteristics or hardware configurations. It is important to recognize that, while the manufacturer may only be supplying the current generation IP, there may be IPs of other generations in circulation. Clinical operation with mixed generations of detectors should be avoided unless it can be shown that variability in the outcome image quality is essentially unaffected.

7.3 X-ray Scatter and Grid Selection

The lower k-edge of the PSP detector using BaFBr(Eu) (k-edge of Ba at 37 keV) may confer a greater sensitivity to scatter than screen-film. Early versions of PSP cassettes did not have adequate backscatter control, which in turn contributed to the formation of artifacts.[16] Apprehension about scatter has caused some practitioners to recommend scatter reduction grids for all bedside exams without regard to patient thickness.

The selection of an appropriate fixed grid for the PSP detector is problematic. There is no universal agreement about what grid type, focused, parallel, or crosshatch; what grid ratio; interspace material; or grid frequency to use. The general-purpose 103-line-per-inch grid presents a periodic signal of about 2 line-*pairs* per mm (4 lines/mm). Since grids are not truly sinusoidal, higher frequency components beyond the fundamental frequency are present, well beyond the Nyquist frequency limit imposed by the detector aperture and sampling pitch. These higher frequency signals can be aliased, causing an appearance of lower frequency grid lines and moiré patterns superimposed on the image, particularly in the slow-scan direction, as "anti-aliasing" signal processing filters cannot be applied. In general, therefore, grid lines should be oriented perpendicular to the scan direction to reduce aliasing. Some manufacturers are recommending high-frequency stationary grids, which are more expensive and difficult to use clinically. At higher frequency, higher ratio grids are required to get equivalent scatter cleanup compared to lower frequency grids, which also have a higher bucky factor (dose penalty).[71] However, in certain situations, grid cutoff is more significant a problem than aliasing. For instance, in applications such as bedside radiography, grid cutoff (absorption of primary radiation) caused by tilted grids can be problematic, particularly for cross-wise imaging of the chest, resulting in very poor image quality. Decubitus grids (grids with lead strips parallel to the short dimension of the grid) can be very useful in reducing grid cutoff for these cross-wise image acquisitions, as positioning of the cassette and grid cap is less critical in the cranial-caudal direction in these situations. In addition, histogram analysis can yield different results with and without grids. Specific menu selections (or processing algorithms) must be considered for grid and non-grid exams to produce optimal results.

Display of a reduced size PSP image on a monitor that eliminates pixel data (e.g., eliminating every other pixel and every other row, instead of downsize averaging) generates moiré artifacts of variable frequency with fixed grids. This is particularly noticeable on thumbnail or reduced size images used by the technologist. Only by displaying the full size image do the moiré artifacts disappear; therefore in these cases it is important to instruct the technologists of this possibility. Grid use in bedside radiography is a particular problem with length-wise chest imaging, where grid cutoff can easily occur when the grid is slightly tilted relative to the x-ray beam central axis. Linear grids oriented in the short axis direction are less vulnerable to the grid cutoff and should be considered for use.

7.4 Radiation Exposure

Current PSP systems tend to require more radiation to produce images of equivalent quality compared to 400-speed rare-earth screen-film systems in common use today. PSP systems are much more tolerant of inappropriate technique than screen-film, and are capable of producing a diagnostic quality image under conditions of under- and over-exposure that would necessitate a repeated examination using screen-film. In other words, in the hands of an experienced user, PSP imaging allows the *choice* of exposure. Tolerance of inappropriate exposure factors with PSP radiography is a double-edged sword: under- and overexposures are not obvious from the appearance of the normalized PSP image. Instead of a light or dark film, reliance on derived indices of exposure based on the results of the normalization process is necessary to monitor effective detector speed, and indirectly patient radiation exposure. These indices differ among manufacturers and are greatly affected by readout and display processing characteristics of the PSP device. There should be a thorough understanding of the exposure indicators and how they relate to estimated (equivalent) detector speed and patient dose (similar to the way estimates for screen-film detectors are determined) as explained in sections 4.2 and 4.3. From a radiation management prospective, it is critical to report the exposure index with the image in any clinical practice of PSP radiography. A method for periodic monitoring of the incident exposure indicator to identify undesirable trends of inappropriate technique (particularly the overexposures, which are more difficult to identify by visual inspection of the image) is essential. The exposure index should be protected from alteration, and audit logs are highly recommended. These capabilities may or may not be available on current systems, depending on the specific equipment and manufacturer, but are considerations for future purchase requirements.

7.5 Phototimer Calibration

In most clinical situations, the primary method of exposure factor control is the phototimer, also known as the automatic exposure control (AEC). Traditionally, phototimers have been designed to provide constant optical densities for a variety of kV and attenuation combinations. This requires that the exposure to the image detector be controlled in a manner that is appropriate for the energy response of the detector in use. It is important to recognize that the response of a PSP detector is different from most conventional screen-film detectors. When setting up a phototimer station for use with a PSP device, the goal should be to produce a constant pixel value for a variety of kV and attenuation combinations. This goal can be met by several methods. (1) If possible, disable the auto ranging during phototimer calibration. Under these conditions, the PSP response (pixel value) can be related to the detector absorbed dose. Phototimer calibration can then proceed in a manner identical to screen-film systems using pixel value (or hard-copy OD when using a laser printer) as the output variable to be adjusted. (2) Calibrate the exposure index response according to manufacturer recommendations (kVp, filtration, etc.). Use the exposure index value as the means to adjust the sensitivity of the phototimer through repeated exposures and measurements of the IPs until the desired exposure index value is achieved.[48] A calibration device introduced in 2004 to perform this type of measurement electronically by emulating the characteristics of the PSP phosphor, monitoring the PSL intensity, and providing an equivalent exposure index value is under study.[72] (3) Patient (phantom) exit exposure can be used as the output variable to be controlled. Calibration of the system using an SNR criterion can be followed.[73]

7.6 PSP System Interfaces to PACS

PSP systems are the foundation for digital projection imaging in the clinical environment. Interfacing a PSP system involves several components, in addition to the PSP reader:

(1) *Image workstation and processing module* contains interfaces to the Radiology Information System (the RIS is a database of patient and examination information, and also provides scheduling capabilities) and to the picture archiving and communications system (the PACS is the repository and electronic distribution system for the digital images generated by the PSP system). With current technology, the dedicated workstation accepts images from the PSP reader using the vendor's proprietary formatting, applies appropriate image processing, and tags the images with the patient demographic information and image formatting, size, and bit depth information among other descriptors.

(2) *Modality work list* interfaces to the RIS and PACS generate information for the technologist indicating the scheduled examination. These interfaces are often facilitated by a PACS "broker," a general-purpose product that interprets the HL7[2] communication data stream from the RIS and produces DICOM Modality Work List (MWL) communication at the ID terminal or workstation to directly input patient demographics associated with the image and examination information. MWL allows the technologist to choose the patient to be imaged by name, accession number (a number generated by the RIS that explicitly points to a specific examination), or medical record number, among other indicators.

(3) The *DICOM interface* to the PACS provides the communications link between the PSP reader, the PSP workstation, the PACS database, and the PACS archive. The interface specifies the digital format of the image information, the unique study information regarding the patient and examination criteria, instructions on communication between the PSP workstation and the PACS archive, and other details that are beyond the scope of this document. During initial installation it is imperative that these interfaces be thoroughly checked against specifications and expectations (e.g., DICOM conformance statements) to ensure the functionality of the system in a PACS environment.

7.7 Technologist Training

When PSP radiography is introduced into a conventional radiography operation, initial technologist training is essential.[74] The technologist must understand the importance of selecting the proper examination, must learn to recognize a new set of artifacts, and must have some idea how to correct inferior images.[75] Appropriate actions with PSP systems are often anti-intuitive to technologists well versed in screen-film radiography. As personnel changes occur, provisions for repeated training should be made, because many personnel will come from a film-based experience.

7.8 Radiologist Acceptance

There are several factors adversely affecting radiologist acceptance of PSP radiography relative to screen-film imaging: (1) Until recently, radiologists were trained to discern clinical features on

film; (2) the American College of Radiology (ACR) teaching file is composed entirely of screen-film images; and (3) for American Board of Radiology (ABR) certification they must demonstrate proficiency at examining screen-film images. PSP images have a different appearance from screen-film; however, when optimally adjusted, PSP systems provide image quality similar to optimal screen-film technology, and also give the ability to interface directly into a digital image network/PACS environment. To ensure optimal adjustments and image quality, acceptance testing and periodic quality control should be implemented.

8. ACCEPTANCE TESTING

Acceptance tests of the PSP system are a first and crucial step toward clinical implementation. Verification of proper function, adherence to functional specifications as published by the manufacturer, documentation of reports, demonstration of personnel training, and establishment of a standard for subsequent quality control tests compose the rationale for these procedures. Several references regarding PSP system acceptance testing are available in the literature.[56,76–83] The reference by Samei et al.[56] uses an early draft of this task group document and provides suggested criteria and tests for quantitative assessment of PSP systems. Also available are spreadsheets authored by Dr. Samei[84] that can be used by the physicist to input data, to analyze results, and to define acceptance criteria as suggested herein, or to define criteria that the physicist can determine independently. Most of the tests described in this task group document are directly applicable to cassette-based, "point-scan" PSP systems. With technological advances, dual-side readout, "cassetteless" PSP systems, and "line-scan systems" are now in the clinical environment. Adaptation to these systems is straightforward, with only minor changes required in the described tests.

8.1 Preliminary Communication with Vendor Engineer/Specialist

Prior to initiating the acceptance test procedures, an outline of the specific tests to be accomplished should be provided to the service engineer during installation. Perusal of applications manuals and other documentation that provides information about the PSP system, instructions for use, and system specifications are extremely important. An itemized list of the components, peripherals, and options delivered with the system should be available. Communication with the service engineer, applications personnel, and sales representative is extremely helpful in getting knowledge about the system, its capabilities, and options.

8.2 Preparations and Initial Adjustments for Acceptance Test Measurements

There are a number of adjustments and calibrations that must occur on the PSP unit prior to acceptance tests and clinical service.[77] These adjustments are typically vendor specific, and should be accomplished in concert with the service engineer at the beginning of the acceptance testing procedures. DICOM conformance statement agreements and MWL functionality (RIS/PACS broker interface via HL7 communication) should be verified for proper operation. PSP acceptance tests do not explicitly describe the tests for the hard-copy laser printer and soft-copy display components, which are also extremely important.

8.2.1 Customization of Alphanumeric Data Recording

Demographic information to be listed on each printed film or displayed image should be reviewed and checked for accuracy. Information includes, but is not limited to, hospital name, machine identification, display processing parameters, exposure index information, date, and time.

8.2.2 Adjustment of the Hard-Copy Recording Device

In situations where primary diagnosis is performed with hard-copy laser films generated from the acquired digital data, the following steps must be checked to ensure proper operation.

(1) The film image is properly positioned on the film.

(2) Shading variations caused by non-uniformities in the laser-light intensity across the film are minimal.

(3) Processor chemistry is maintained at an optimal level.

(4) Internal laser calibration is within tolerance limits specified by the vendor.

There will typically be system-generated test scans available to assist in the verification of these parameters. In the absence of an appropriate internal test scan, a test pattern (such as TG18-QC[70] or SMPTE*[85]) may be printed from a PACS workstation to verify printer calibration stability.

8.2.3 Film Processor and Laser Printer Tests

A film processor audit requires the verification of proper chemistry activity, replenishment levels, developer temperature, and lack of processor-generated artifacts for wet-chemistry systems. The film processor should be evaluated according to the manufacturer's quality assurance recommendations as well as methods outlined in the literature.[86–88] The film processor chemistry and the developer temperature influence the OD values of the printed films. It is advisable to independently test processor performance with sensitometric strip methods on a daily to weekly basis, depending on system use. This will allow any potential problems with the film processor to be uniquely identified and separated from those caused by the PSP system hardware misadjustment.

Dry-laser systems are rapidly supplanting wet-chemistry systems in the clinical environment. These self-processing systems do not generate any chemical waste, as the film undergoes internal "self-processing."[89] Stability of the dry laser systems and a lack of chemical waste disposal are distinct advantages over wet laser systems, even though the film costs may be somewhat higher.[90]

8.2.4 Laser Printer Calibration and LUT Parameters

Laser printer calibration is typically performed with a built-in sensitometer to render a range of OD steps that are measured with a calibrated densitometer. In some printers, the densitometer is also part of the unit and allows automatic calibration adjustment. With acceptance testing, the

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*Society of Motion Picture and Television Engineers.

laser printer should be tested in a "manual" mode using a calibrated densitometer with the laser-generated sensitometry strip and compared to the "automatic" methods. Should the values fall outside the recommended density ranges as specified by the manufacturer (typical values are $D_{min} \pm 0.03$; low density of ~0.45 ± 0.07; mid density of ~1.20 ± 0.15; high density of ~2.20 ± 0.15), a correction (calibration) algorithm should be invoked on the film laser printer (often under the guidance of the service engineer) and the test repeated. For a second failure, a request for repair should be initiated prior to doing other acceptance tests. In certain situations, the laser printer can compensate for variations of film processor chemistry or malfunction that ultimately leads to a catastrophic failure (particularly with wet-chemistry laser printers). It is important, there-fore, to determine the degree of compensation implemented by the laser subsystem when at all possible.

Ensuring consistent image appearance on the film image and on the image display workstation has historically been a difficult process, requiring numerous printer LUTs resident on the laser printer system that are designed for specific modalities, manufacturers, and image presentations, all of which must be specifically chosen on a case-by-case basis. The physicist must verify that the appropriate image presentation is obtained with a printer LUT that closely matches the appearance of information content on a *calibrated PSP workstation monitor* (see section 8.2.5), which often requires interaction with the printer manufacturer for selection and adjustment. In a PACS environment, the best method of ensuring consistent image presentation is to calibrate all printers (and display monitors) to the DICOM grayscale standard display function (GSDF),[61,91] and directly use presentation p-values to drive the printer and to render grayscale values in a consistent manner.

8.2.5 Image Workstation Display Monitor Calibration/Resolution Tests

In a soft-copy environment, display workstations are a critical link in the overall image quality verification of a PSP system. A short list of items that should be evaluated initially and frequently via QC tests are:

- Adjustment of monitor performance by calibrating the monitor to the DICOM grayscale standard display function (GSDF) and verifying performance by following the AAPM TG 18 document[59,70]

- Calibration of the display LUT and its conformance with DICOM PS3.14[92]

- Determination of high contrast spatial resolution, both centrally and peripherally

- Determination of geometric distortion, particularly in the periphery of the image

- Evaluation of luminance output with a luminance meter

- Evaluation of room lighting conditions with a luminance meter.

Hardware and software tools to perform these tests should be specified in the original purchase agreement to be delivered with the system. This is an area critically important to image display and evaluation of the PSP system. Specific recommendations for soft-copy display acceptance testing and quality control are provided in AAPM task group 18 documentation.[59,70]

8.2.6 Evaluation of PSP System Interfaces: RIS and PACS

Prior to specific acceptance tests, the interfaces to the RIS for DICOM Modality Work List (MWL) and PACS for image storage and electronic distribution should be tested. Communication

with the installation engineer and/or applications specialist during installation and configuration of the system is the best way to verify the interfaces. Correct transfer of patient demographic information, date of birth, accession number, medical record number, examination type, referring physician, site description, date, time, and other informational items must be verified in the meta-data contained in the DICOM header. Although the majority of manufacturers use the CR infor-mation object definition (IOD), it is superseded by the DICOM "DX" IOD, with a richer description of the image details and many more mandatory elements than the "CR" IOD.[93] Image information sent to the PACS should have appropriate image processing (e.g., "For Presentation") or raw, non-processed "For Processing" image data), proper default window/level information and grayscale rendition (e.g., "value of interest (VOI) LUT"), correct image size and orientation (as sent from the local workstations), appropriate transfer speed, and other details as specified in the purchase agreement. This involves cooperation with personnel who oversee other aspects of the PACS, including Information Systems and PACS managers as appropriate.

8.2.7 Characterization of the X-ray Beam

X-ray beam characterization and reproducibility is important for testing sensitivity, linearity, and uniformity of PSP system response. A beam produced by a high-frequency generator system is preferred because of its excellent stability and accuracy. For general-purpose PSP systems, a standard 80-kVp beam should be used. Most manufacturers recommend the addition of 0.5 to 1.0 mm Cu to the beam in order to simulate the removal of lower energy photons by the patient in the transmitted x-ray beam, and to make the beams from different x-ray systems more con-sistent. Consult the specific manufacturers' recommendations for beam filtration; if a different beam filtration is used, ensure that the system initially meets the specifications of the manufac-turer in terms of calibration points before recalibrating the unit. (Note: AAPM TG 116 is rec-ommending the use of a sandwich filter of 1 mm Al, 1 mm Cu, 1 mm Al).[51] An identical beam, or nearly identical beam, should be a part of all subsequent QC testing and performance moni-toring of the PSP system. Also, at the time of acceptance testing, identification of several image detectors to be set aside and safeguarded for the technologist's and physicist's sole use in QC testing is highly recommended. For specialty PSP systems, more appropriate beams should be characterized and used (e.g., 110 to 120 kVp for a dedicated chest PSP system, and 25 kVp for a dedicated mammography PSP system), with appropriate added filtration (at the discretion of the physicist).

Sensitivity and linearity tests should challenge PSP system response under conditions representative of the clinical environment. To assist reproducibility, a filter of 1 mm Cu + 1 mm Al is recommended for general diagnostic CR. The filter should be placed at the collimator with the copper side facing the x-ray tube. Image uniformity tests may be conducted with or without the filter.

8.3 Tools and Equipment Required for Acceptance Evaluation

Table 1 lists the minimum requirements of tools and equipment for acceptance evaluation.

8.4 Specific Testing Procedures

Recommended acceptance tests are listed in Table 2.

Table 1. Recommended Equipment and Tools for Performing PSP Acceptance Tests

- Calibrated x-ray source
- Calibrated hard-/soft-copy display devices
- Densitometer for hard-copy film evaluation
- Copper filters and aluminum filters
- Calibrated ion chamber and test stand
- Manufacturer-approved screen cleaning solution/cloths
- Two metric-calibrated 30-cm steel rulers (laser jitter and distance accuracy)
- High-contrast resolution line-pair phantoms, 4° sector type (up to 5 lp/mm)
- Low-contrast phantom (e.g., Leeds TO.12,[94] UAB phantom,[95] homemade)
- Manufacturer-recommended PSP phantom for periodic quality control testing*
- Screen-contact wire mesh pattern
- Small lead block, ~5 cm × ~5 cm × ~0.3 cm thick for erasure thoroughness test
- Antiscatter grid (10:1 or 12:1, ~100 line/inch)
- Spacer blocks (4) of 5 cm × 5 cm × 20 cm height (to raise imaging cassettes off of floor)
- Lead apron or lead sheet, 35 × 43 cm (to control backscatter)
- Anthropomorphic phantoms (foot, hand, pelvis, chest, if available)
- Timer (stop watch), measuring tape, flashlight, tape
- ≥10X magnification loupe with 0.1 mm graticule

*Use the manufacturer QC phantom(s) to verify manufacturer-provided equipment specifications and to establish baseline
 values for quality control testing.
 UAB: University of Alabama Birmingham.

Table 2. Recommended Testing Procedures for PSP Systems

1. Component and Imaging Plate Physical Inspection and Inventory
2. Imaging Plate Dark Noise and Uniformity
3. Exposure Indicator Calibration
4. Linearity and Auto-ranging Response
5. Laser Beam Function
6. Limiting Resolution and Resolution Uniformity
7. Noise and Low-Contrast Resolution
8. Spatial accuracy
9. Erasure Thoroughness
10. Aliasing/Grid Response
11. IP Throughput
12. Positioning and Collimation Errors

Many of these tests require x-ray exposure to the IP in a known and reproducible manner. A recommended setup is illustrated in Figure 21. A calibrated x-ray source with reproducible output should be employed and a source-image distance (SID) of at least 180 cm (to minimize beam divergence and heel effect variations) is desirable, with the central axis centered to the IP. Recommended is the use of a lead attenuator (lead apron or lead sheet) to minimize backscatter and spacers on which the IPs are above the floor at the 180-cm distance. X-ray tube collimators should be adjusted with at least a 5-cm margin outside of the IP. Measurement of the output exposure is first determined along the central axis of the x-ray beam free-in-air, above the IP, at about 125-cm source-chamber distance (SCD), and corrected by inverse square falloff to determine the incident exposure on the IP. Adjustment of the kVp and mA (with any additional added filtration, as required) to achieve a known exposure is then determined. Check initial reproducibility with five exposures; the coefficient of variation (COV) should be less than 0.05, otherwise use a more reproducible x-ray tube/generator. The chamber is moved to the periphery of the field (in the beam but outside the central area of the IP), and five repeat exposures are made to give an "off-axis ratio" measurement relative to the central position. Ensure the COV < 0.05, and use the ratio to correct for the exposures along the central axis. Measured exposure values at the periphery of the field are converted to incident exposure to the IP as:

Central IP exposure = measured peripheral exposure $\times$ (SID/SCD)^2 $\times$ off-axis ratio.
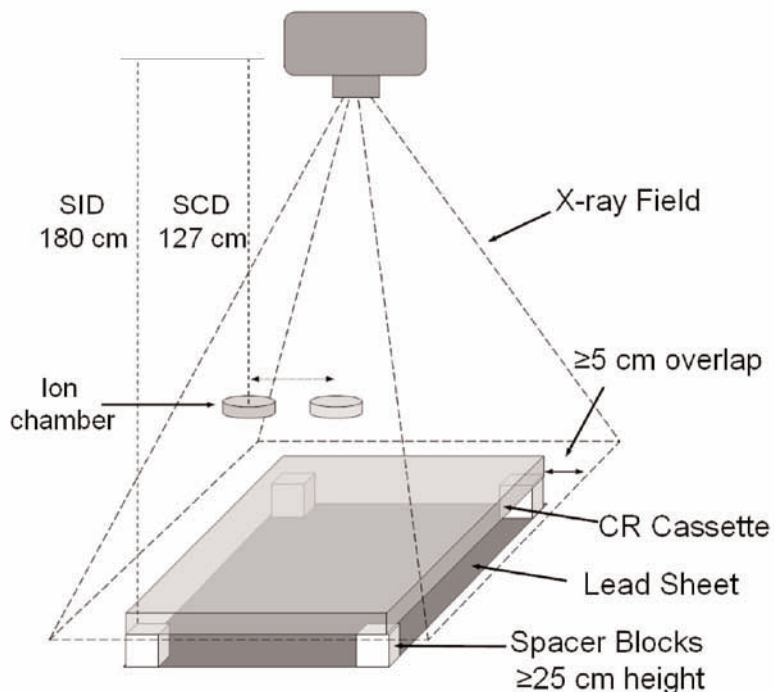

Figure 21. Recommended acquisition geometry for exposing PSP IPs. See text for details on ion chamber position. The SCD of 127 cm is arbitrary (gives a reasonable distance from the focal spot and the detector) and results in an inverse square correction factor of 0.50.

NOTE: In Tables 3–13, there is no intended preference given to any manufacturer. At the time of the original work of the task group, these were the manufacturers with the most systems deployed clinically, from which specifications and results were derived. Appendix A lists manufacturers available at the time of publication. In addition, the pixel value (PV) and the pixel value standard deviation (PVSD) are the average and the standard deviation of the values within the ROI defined by the user. For some manufacturers, these values are not readily obtainable, which requires analysis of the images on a separate computer system independent of the QC workstation. [Tables 3–13 in this report are adapted from Tables III–XV in reference 56: E. Samei, J. A. Seibert, C. E. Willis, M. J. Flynn, E. Mah, and K. L. Junck. "Performance evaluation of computed radiography systems." *Med Phys* 28:361–371 (2001), with permission from AAPM.]

8.4.1 Component Inventory

Imaging plates, cassettes, hardware, and associated documentation delivered with the system should be inventoried and inspected. Items to inspect include the proper installation of the main unit, the processor, power lines, exhaust ducts, water supply, developer/fixer replenishment tanks, hose connections, and environmental air conditioning. The phosphor plates are particularly vulnerable to mishandling. A careful, visual inspection of each plate is necessary. Surface defects or scratches are noted as found, and logged on the inventory checklist with the corresponding serial number. Each cassette should also be examined for loose or protruding screws or fasteners. Include all findings in the final acceptance test report.

8.4.2 Imaging Plate Dark Noise and Uniformity

All IPs in the inventory must first be erased with the full erasure cycle to ensure removal of all residual signals from background radiation or other sources. The erasure unit subsystem is typically composed of a high-pressure sodium or fluorescent lamp (this depends on the manufacturer and model number). After erasure, several plates (e.g., three to five) should be scanned using an automatic scaling algorithm or fixed scaling algorithm to drive the gain of the system to maximum. On some systems, a "dark-current" situation causes the automatic adjustment of the readout technique to wide latitude, nominal exposure, with little or no signal amplification. In this case, use a fixed manual technique and drive the system to a high amplification signal. Testing parameters are listed in Table 3 for the dark noise tests for the three major manufacturers of PSP units. The resultant soft- or hard-copy image for each IP should demonstrate a clear, uniform, artifact-free image when viewed with clinical window width and level settings. Exposure indicators (for automatic processing) should have a 0 (or null) exposure value. Obvious artifacts, density shading, or non-uniformities present on any output image should be evaluated further. If more than two plates of the test set have a problem, all of the plates in the inventory should be tested. Reproducible artifacts on a number of images or films, such as a uniform shading response, indicates the laser subsystem, light collection guide, memory board, erasure unit, or fogged film are potential problems (assuming the printer has been evaluated for uniformity prior to the test). Corrective action is required before proceeding to other tests.

Table 3. Testing and Acceptance Criteria for Dark Noise Evaluation

	Agfa	Fuji	Kodak
Exposure	No exposure to IP. Use freshly erased IPs.		
Processing	System diagnosis flat field, speed class = 200	Test/sensitivity (L = 1), fixed EDR (S = 10,000)	Pattern
Image post-processing	None Musica parameters = 0.0 Sensitometry = linear	Linear (GA = 1.0, GT = A, RE = 0.0)	Raw data, no edge enhancement settings, window = 512, level = exposure index
Measurements	lgM, PV, PVSD, scan average level (SAL)	PV and PVSD	Exposure Index (EI), PV, PVSD
Qualitative criteria	Uniform image, no artifacts	Uniform image, no artifacts	Uniform image, no artifacts, except profile bands in IP direction
Quantitative criteria ROI over 80% of image	lgM < 0.28 SAL < 130 PV < 350 PVSD < 5	PV < 280 or PV > 744 for inverse grayscale PVSD < 4	$EI_{GP} < 80$ $EI_{HR} < 380$ $PV_{GP} < 80$ $PV_{HR} < 80$ $PV_{SD} < 4$

Note: PV = average pixel value; PVSD = average pixel value standard deviation within the defined region of interest; HR = High resolution IP, GP = general purpose IP for the Kodak PSP, EDR = exposure data recognizer. [Adapted from Table III in reference 56 with permission from AAPM.]

Image uniformity verifies appropriate response of the IPs to a high incident exposure (~10 mR, 80 kVp, 0.5 mm Cu and 1 mm Al, 180 cm SID) to reveal variations in x-ray response. Make certain that the high exposure does not saturate the ADC response (i.e., all pixels have the maximum value, for example, 4095). If so, repeat with an incrementally lower exposure until an appropriate image is obtained. *Each* cassette/phosphor plate in the inventory is centered to the x-ray beam and uniformly exposed. A reproducible geometry and plate orientation must be maintained. If significant heel effect variation is present, two sequential half exposures with the cassette orientation rotated by 180° are necessary. This test is applied to all IPs in the inventory. Table 4 provides guidance for specific tests; consult the manufacturer for specific settings for acquisition and processing.

For film output (hard-copy), optical densities are measured in the center of each quadrant of the film and in the center position to determine absolute density and spatial uniformity. Central film density is acceptable if within ±0.10 OD of the programmed OD value (usually 1.20). Spatial uniformity is acceptable when all measured OD values are within ~10% of the average OD. For soft-copy evaluation of the images on a workstation, the average digital value of each ROI should be within 10% of the global average. Standard deviation should also be similar in each of the five ROIs.

Table 4. Testing and Acceptance Criteria for IP Uniformity Response

	Agfa	**Fuji**	**Kodak**
Exposure	Use 10 mR incident exposure at 80 kVp with 0.5 mm Cu and 1 mm Al filtration and 180 cm SID.		
Processing	System diagnosis flat field, speed class = 200	Test/sensitivity (L = 1), Semi EDR	Pattern
Image post-processing	None Musica parameters = 0.0 Sensitometry = linear	Linear (GA = 1.0, GT = A, RE = 0.0)	Raw data, no edge enhancement settings, Window = 512, Level = exposure index
Measurements	PV and PVSD	PV and PVSD	PV and PVSD
Screen to screen variation	Standard deviation of lgM (LMSD) and mean/standard deviation of PV among screens (PVs and PVSDs)	Standard deviation/mean sensitivity (SD/Ss) and mean/standard deviation of PV among screens (PVs and PVSDs)	Standard deviation of Exposure Index among screens (EISDs)
Qualitative criteria	Uniform image without any visible artifacts with window/level adjustments		
Quantitative criteria ROI over 80% of image	PVSD < 25 single screen LMSDs < 0.02 PVSDs < 25	PVSD < 20 single screen SD/Ss < 5% PVSDs < 20	PVSD < 20 single screen EISDs < 20

[Adapted from Table IV in reference 56 with permission from AAPM.]

For either hard-copy or soft-copy image evaluation, all images should be examined for banding, black or white spots, and streaks. Scan line dropouts are detectable as lucent straight lines and represent dust/dirt particles on the pickup light guide, a fairly common artifact. "Unique" artifacts are usually traced to the IP in question, while artifacts appearing consistently on several or all images are likely due to the equipment (reader or writer components of the system). For artifacts identified to a specific IP, initiate plate cleaning followed by primary erasure, and then retest. If the problem(s) still exist(s), the IP should be removed from service. In the case of a consistent variation in OD shading across the film, exposure of the same plate in 180° orientations will cancel any possible variations due to the x-ray tube heel effect. If the shading variation still persists after this action, the service engineer should implement a shading correction calibration. The presence of image artifacts indicates suboptimal performance and necessitates corrective action by service personnel.

8.4.3 Exposure Indicator Calibration Accuracy

The exposure indicator is a method to determine a surrogate measure of the PSP detector equivalent radiographic speed for a given exposure. Incident exposure to the plate of ~1 mR is used to establish "exposure index" accuracy. A time delay between the exposure and readout (e.g., 10

minutes) is required by some manufacturers to reduce variation in phosphorescence lag, while others do not have such a requirement. The task group consensus recommends a 10-minute delay. Additionally, there is no standardization of beam quality between manufacturers, as each has a specific kVp and tube filtration that are required for their system calibration (consult with the specific manufacturer for recommendations). A standard x-ray beam filtration with 0.5 mm Cu and 1 mm Al is recommended by the task group, as variations in exposure index are reduced with a more filtered beam.[82] More recently, AAPM TG 116 on standardization of dose index for digital radiography (in progress, October 2005)[50] has specified an attenuation filter of 1 mm Al, 1 mm Cu, 1 mm Al in a sandwich configuration for pre-hardening the indecent beam used for Exposure Index calibration. Ultimately (and hopefully within the next several years), establishment of a standardized method to estimate the incident exposure and system speed for a specified beam quality will occur for all DR systems. Table 5 lists testing and acceptance criteria for exposure indicator accuracy.

Table 5. Testing and Acceptance Criteria for Exposure Indicator Accuracy

	Agfa	**Fuji**	**Kodak**
Exposure: Manufacturer specified	1 mR @ 75 kVp, 1.5 mm Cu filtration, no delay to readout	1 mR @ 80 kVp, no filtration, 10 minutes to readout	1 mR @ 80 kVp, 0.5 mm Cu + 1 mm Al, 15 minutes to readout
Processing	System diagnosis flat field, speed class = 200	Test/sensitivity (L = 1), semi EDR	Pattern
Image post-processing	None Musica parameters = 0.0		
Measurements	lgM and lgM normalized to 1 mR screen exposure ($lgM_{1 mR}$) using $lgM_{1 mR} = lgM - log(exp)$, SAL normalized to 1 mR screen exposure ($SAL_{1 mR}$) as $SAL_{1 mR} = SAL/(exp)^{0.5}$	Sensitivity and sensitivity normalized to 1 mR exposure to the screen ($S_{1 mR}$) using $S_{1 mR} = S_{exposure}$	Exposure index (EI) and exposure index normalized to 1 mR exposure to the screen ($EI_{1 mR}$) using $EI_{1 mR} = EI -1000 \times log(exposure)$
Qualitative criteria		None	
Quantitative criteria	$lgM_{1 mR}$ 2.2 < ±0.045, single screen $lgM_{1 mR}$ 2.2 < ±0.023, all screens averaged $SAL_{1 mR}$ 1192 < ±60, single $SAL_{1 mR}$ 1192 < ±30, all screens averaged	$S_{1 mR}$ 200 < ±20, single screen $S_{1 mR}$ 200 < ±10, all screens averaged	$EI_{1 mR}$ 2000 < ±45, single screen $EI_{1 mR}$ 2000 < ±23, all screens averaged

[Adapted from Table V in reference 56 with permission from AAPM.]

8.4.4 System Linearity and Autoranging Response

This test determines the response of the detector and readout systems to at least three decades of exposure variation (1000 times difference). A calibrated radiographic x-ray tube with reproducible output (kV accuracy better than ±5% and exposure output accuracy ±2%) and acquisition geometry/detector orientation must be maintained. Suggested techniques are 80 kVp, 180 cm SID, and 0.5 mm Cu + 1 mm Al filtration, with the beam collimated just outside the total detector area. (The aluminum filter is placed towards the detector and eliminates any possible characteristic radiation emanating from the copper.) Determine radiographic techniques to provide incident exposures of approximately 0.1, 1.0, and 10 mR. On some PSP readers, a 10-mR exposure may saturate the ADC. If so, incrementally reduce this exposure until a nonsaturated result is achieved. Actual incident exposure should be measured with a calibrated ionization chamber free-in-air (no backscatter) and calculated to the surface of the PSP detector. For each incident exposure, acquire three independent images, and use a fixed delay time of 10 minutes between exposure and processing.

Table 6 lists the processing, measurements, and evaluation criteria. If using hard-copy film as the output, the relationship between pixel value and OD should be established beforehand using an electronic test pattern, and then should be incorporated as a transformation in the quantitative analysis of the results. Because the beam filtration does not conform to the Agfa and Fuji recommendations, the lgM and S numbers might not give the exact calibrated exposure indicator response expected. An option is to use the manufacturer's recommended filtration, as this is a relative test. In some cases without filtration, however, it is difficult to achieve low incident exposures (e.g., 0.1 mR) at the specified geometry. Repeat the process a total of three times (a total of nine images), being careful to use the same PSP detector for a specific incident exposure measurement. Calculated exposure values should be within ±20% of the actual incident exposure for any single detector and within ±10% for the average. Qualitative image noise characteristics in the resultant film images should be inversely related to incident exposure. Resultant OD of each film should be within ±0.1 OD of the programmed value. Quantitative evaluation of the image properties on a computer workstation should verify a consistent average digital number independent of exposure, and a decrease in relative noise (increase in signal to noise) with increased exposure. The quantum-limited operation range is determined by plotting the standard deviation of the noise relative to the log incident exposure and determining the linear fit of the line with a slope of 0.5. Deviations from a straight-line response and a slope not equal to 0.5 indicate other noise sources that are interfering with the quantum-limited operation of the PSP system.

8.4.5 Laser Beam Function

Laser beam scan line integrity, beam jitter, signal dropout, and focus are evaluated in this test. Use a radiographic technique of ~60 kVp, 180 cm SID, and mAs to deliver an incident exposure of ~5 mR. Place the steel ruler on a 35 × 43 cm (14 × 17 in.) centered on the cassette and nearly perpendicular to, approximately 5° from the laser beam scan lines. Table 7 summarizes the exposure conditions, IP processing/post processing details, and the qualitative/quantitative criteria to determine adequate system function.

Laser beam jitter (inconsistent gray-level output caused by timing errors with the location of the beam or synchronization with the ADC) is evaluated by examining the edge of the ruler on the image. Ruler edges should be straight and continuous over the full length of the hard-copy or soft-copy image. Scan lines in light to dark transitions along the ruler edge that do

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Table 6. Testing and Acceptance Criteria for Linearity and Autoranging Response

	Agfa	Fuji	Kodak
Exposure condition	Use approximate 0.1, 1.0, and 10 mR incident exposures at 80 kVp with 0.5 mm Cu and 1 mm Al filtration (or the manufacturer's recommended filtration for exposure index value calibration) and 180 cm SID. Use a consistent delay time between exposure and readout.		
IP processing	System diagnosis flat field, speed class = 200	Test/avg 4.0; Semi EDR & Fix EDR = 200; repeat using test/contrast with same EDR parameters	Pattern
Image post-processing	None Musica parameters = 0.0	Linear GA = 1.0, GT = A, RE = 0.0	Raw data and no edge enhancement settings
Measurements	lgM, avg pixel value (PV), and scan avg level (SAL) over 80% of image area. Slopes and correlation coefficients (CCs) of linear fits to log(SAL) vs. log(E), PV vs. log(E), and lgM vs. log(E).	*For semi EDR,* correlation coefficient (CC) of a linear fit to log(S) vs. log (E) plot. *For fixed EDR,* avg pixel value (PV) within 80% of the image area, slope and correlation coefficient (CC) of a linear fit to PV vs. log(E).	Exposure index (EI) and avg pixel value (PV) within 80% of the image area. Slope and correlation coefficient (CC) of a linear fit to EI vs. log(E) and PV vs. log(E) plots.
Qualitative criteria	SAL vs. exposure on a linear-log plot should result in a straight line.	*For semi EDR,* slope and correlation, sensitivity vs. exposure on a log-log plot should result in a straight line. *For fixed EDR,* to PV vs. exposure on a linear-log plot should result in a straight line.	The plot of EI and PV vs. exposure on a linear-log scale should result in straight lines.
Quantitative criteria	Slope$_{lgM}$ – 1 < ±0.1 Slope$_{SAL}$/0.5 – 0.1 < ±0.1 Slope$_{PV}$/1250 – 0.1 < ±0.1 CCs > 0.95	Slope$_S$ + 1 < ±0.1 Slope$_{PV}$/256 – 1 < ±0.1 (Avg 4)[a] Slope$_{PV}$/511 – 1 < ±0.1 (Con)[a] CCs > 0.95	Slope$_{EI}$/1000 – 1 < ±0.1 Slope$_{PV}$/1000 – 1 < ±0.1 CCs > 0.95

[a] In some Fuji systems, an inverse relationship exists between PV and log(E), and the polarity of the equation should be reversed.

[Adapted from Table VI in reference 56 with permission from AAPM.]

Table 7. Testing and Acceptance Criteria for Laser Beam Function

	Agfa	Fuji	Kodak
Exposure condition	Place a steel ruler perpendicular to the laser scan direction. Use 5 mR incident exposure at 60 kVp with no added filtration.		
IP processing	System diagnosis flat field, speed class = 200	Test/sensitivity Semi EDR	Pattern
Image post-processing	None Musica parameters = 0.0 Sensitometry = linear	Linear GA = 1.0, GT = A, RE = 0.0	Raw data and no edge enhancement settings Window = 512 Level = exposure index
Measurements	Examine edges of ruler on soft-copy display using 5 to 10× magnification; if on film, use 10× magnification loupe; if any jitter present, obtain jitter dimension with workstation measurement or ROI tool.		
Qualitative criteria	Ruler edges should be straight and continuous without under- or overshoot of scan lines in light to dark transitions.		
Quantitative criteria	There should not be more than occasional ±1 pixel jitters over the ruler edges.		

[Adapted from Table VII in reference 56 with permission from AAPM.]

not form the linear ruler edge indicate a timing error, or laser beam modulation problem. View the image scan lines with a 5× (or greater magnification) in various areas across the image to check for uniform spacing.

8.4.6 Limiting Resolution and Resolution Uniformity

Spatial resolution tests include measurement of the central and peripheral limiting resolution for each IP size and type (standard and high resolution) along the scan and sub-scan directions. Place three line pair patterns (or as available) on the IP, two along the scan and sub-scan directions, and a third at 45°. Expose each to ~5 mR using 60 kVp and unfiltered beam at 180-cm SID. To determine consistency of the resolution response across the IP, use a fine wire mesh pattern (e.g., mammography screen-film contact tool). Table 8 summarizes the exposure conditions, processing, and evaluation of the results.

Use a readout/processing algorithm to enhance radiographic contrast *without* significant edge enhancement. For a soft-copy display device, zoom the digital image to the intrinsic resolution limit, and adjust window/level for best visualization of the object. Both central and peripheral resolution should indicate a response close to the maximum resolution specified for the individual combination of reading sampling rate and phosphor type. Often, the resolution in the scan direction will be less than the resolution in the sub-scan direction due to the PSP lag response and/or the implementation of an anti-aliasing filter algorithm, which results in blurring of the signals. If the spatial resolution is not within 10% of that indicated in the manufacturer's specifications for either the vertical or horizontal directions, corrective action should be initiated. Measured resolution can exceed the theoretical sampled resolution limit if the test pattern is

Table 8. Testing and Acceptance Criteria for Resolution and Resolution Uniformity

	Agfa	Fuji	Kodak
Exposure condition	Use a 5 mR incident exposure with an unfiltered 60 kVp beam at 180 cm SID. High contrast resolution test patterns measure central resolution of the IP. The mesh pattern measures resolution uniformity across the IP.		
IP processing	System diagnosis flat field, speed class = 200	Test/sensitivity Semi EDR	Pattern
Image post-processing	None Musica parameters = 0.0 Sensitometry = linear	Linear GA = 1.0, GT = A, RE = 0.0	Raw data and no edge enhancement settings Window = 512 Level = exposure index
Measurements	Maximum spatial resolutions perceived in each direction (R_{hor}, R_{ver}, $R_{45°}$) using ~10× magnified view of images with narrow window. On film, verification with 10× loupe.		
Qualitative criteria	Wire mesh image should be uniform and free of any blurring across the image.		
Quantitative criteria		$R_{hor} / f_{Nyquist} > 0.9$ $R_{ver} / f_{Nyquist} > 0.9$ $R_{45°} / (1.41 f_{Nyquist}) > 0.9$	

[Adapted from Table VIII in reference 56 with permission from AAPM.]

positioned at a diagonal to the x-y matrix. In this case, the "effective" sampling (pixel) pitch of the resolution pattern is smaller by the sine of the angle (e.g., 0.707 for 45° angle), thus exceeding the actual limiting vertical or horizontal resolution. These tests are subjective and prone to error, but are usually sufficient for verification of appropriate spatial resolution response. Note that a potential problem with the use of a bar phantom is the possibility that the high-frequency bars will be represented at a lower frequency due to aliasing. The observer may believe that the bars are seen correctly, when in fact they are not, which can lead to an erroneous estimate of limiting resolution. A more comprehensive and quantitative evaluation of a PSP system is achieved by measuring the modulation transfer function (MTF).[96] Many manufacturers are implementing MTF measurements in their periodic QC packages (software and hardware).[97]

8.4.7 Noise and Low-Contrast Resolution

Contrast resolution should be limited by quantum statistics (random variations in the number of x-rays absorbed in the IP) in a well-designed system. A specific low-contrast phantom (or a design) was not determined by the task group, but a calibrated low-contrast test object such as the Leeds phantom designed for computed radiography[83,94] or the UAB low-contrast phantom[95] are appropriate for use, as are others. For the Leeds phantom setup, 75 kVp with 1 mm added Cu filtration is used with a standard clinical acquisition protocol (e.g., contact imaging with a grid; PSP cassette placed in a table bucky, etc.). In Table 9, acquisition procedures for the Leeds TO.12 is listed. Three individual images are acquired with incident exposures to the imaging plate of ~0.5 mR, ~1.0 mR, and ~5 mR.

Table 9. Testing and Acceptance Criteria for Noise and Low-Contrast Resolution

	Agfa	Fuji	Kodak
Exposure condition	Perform this test for each size IP. Use a low-contrast phantom; a Leeds TO.12 is used as an example here. Three images of the phantom are acquired at 0.5, 1.0, and 5.0 mR incident to the IP using 75 kVp and 1 mm Cu filtration. Use a delay to readout of 10 minutes after exposure.		
IP processing	System diagnosis flat field, speed class = 200	Test/sensitivity Semi EDR	Pattern
Image post-processing	None Musica parameters = 0.0 Sensitometry = linear	Linear GA = 1.0, GT = A, RE = 0.0	Raw data and no edge enhancement settings Window = 512 Level = 4096 – EI (GP) Level = 3796 – EI (HR)
Measurements	Minimum discernible contrast for each object size (contrast detail threshold), standard deviation of Pixel Value (PVSD) within a fixed small region of the images, correlation coefficient (CC) of linear fit to log(PVSD) vs. log(E).		
Qualitative criteria	Contrast-detail threshold should be proportionately lower at higher exposures.		Contrast-detail threshold should be proportionately lower at higher exposures, with higher contrast thresholds for GP IPs.
Quantitative criteria		CC > 0.95[a]	

[a]The presence of scatter might render the quantitative results less valid, as there is an assumption of a logarithmic relationship between pixel value and exposure. (See Table 15.)
[Adapted from Table IX in reference 56 with permission from AAPM.]

An alternative method to determine quantum noise-limited operation is to extend the measurements in section 8.4.4 to include a wider range of incident exposures (e.g., 0.1, 0.3, 0.5, 0.7, 1.0, 2.0, 5.0, and 10.0 mR) whereby multiple exposures properly collimated on a single plate can be obtained. Transferring the image to an independent digital workstation (if necessary), the standard deviation within a ROI over a uniform area for each exposure is measured, and the variance (std dev)^2 versus the reciprocal of the incident exposure to the detector is plotted. A linear least-squares fit to the data (excluding any obvious outliers which indicate non-quantum limited statistics) should have $r^2 > 0.98$ and a slope of ~1.8 – 2.0, within an incident exposure range of ~0.2 to 5.0 mR. Note that these performance criteria are based only on one system (Fuji 5000) and are not necessarily applicable to other systems.

A more robust measurement of noise is obtained by measuring the noise power spectrum (NPS), noise equivalent quanta (NEQ), and detective quantum efficiency (DQE) as a function of spatial frequency at various exposure levels.[20,52,55,98,99]

Contrast sensitivity should improve with increased exposure in the clinically relevant range, otherwise something is amiss. Other sources of noise and factors should be considered, including fixed point noise (artifacts), excessive luminance or amplification noise, and x-ray/light scatter influences on the subject contrast.

8.4.8 Spatial Distance Accuracy

Spatial distance accuracy is determined with "x-ray" ruler lead markers or from flat objects with known dimensions such as a resolution bar phantom. In addition, the accuracy should be maintained from the center to the periphery along both image axes. A periodic uniform grid pattern is also useful to verify a uniform pattern over the whole image without distortion. For soft-copy display, the measurement of distance accuracy using electronic calipers on the projected x-ray ruler is straightforward; however, for film images, observe magnification, M, of the *printed* image (which can increase: M>1, or decrease: M<1 the image size). Laser-generated films will sometimes reduce the image size a slight amount to compensate for image borders (e.g., M = 0.95 or 5% smaller image size). Table 10 summarizes the distance accuracy tests, including processing and evaluation steps.

Pixel calibration verification is initially performed for each IP/image matrix size as clinically used. The electronic distance calipers are used on the x-ray ruler image to measure distance accuracy in the horizontal and vertical directions directly, and is best performed over distances of 15 cm or greater when possible for the soft-copy images. Recommended is the use of image zoom to enable accurate and reproducible placement of the calipers. If image magnification is present, the estimated *actual* distance is determined by dividing the measured distance by M. If images are to be printed, spatial accuracy should also be evaluated for each print mode used clinically (for example, best fit, true size, 2:1, etc.). In some cases, workstations provide internal pixel calibration by measuring known object dimensions and updating the pixel dimensions accordingly, which also should be tested. Comparison of the true distance to the measured distance corrected for magnification should be within measurement error, and less than 2%, but

Table 10. Testing and Acceptance Criteria for Spatial Accuracy

	Agfa	**Fuji**	**Kodak**
Exposure condition	Place a regular wire-mesh contact test tool on a cassette of each size. Expose the cassette to ~5 mR using a 60 kVp beam without filtration at 180 cm SID. Repeat with another IP of the same size using "x-ray" rulers in the vertical and horizontal directions.		
IP processing	System diagnosis flat field, speed class = 200	Test/sensitivity Semi EDR	Pattern
Image post-processing	None Musica parameters = 0.0 Sensitometry = linear	Linear GA = 1.0, GT = A, RE = 0.0	Raw data and no edge enhancement settings Window = 512 Level = exposure index
Measurements	Measure distances in the orthogonal directions over 15 cm (or greater), using the measurement tool on the QC workstation; on film, directly measure the ruler, taking into consideration any possible image minification/magnification.		
Qualitative criteria	Grid pattern spacing should be uniform without any distortion across the image.		
Quantitative criteria	Measured distance should be within 2% and preferably within 1% of actual.		

[Adapted from Table X in reference 56 with permission from AAPM.]

preferably less than 1% *in both directions*. Where possible, the use of a larger distance (e.g., 25 to 30 cm) reduces the percentage of the measurement errors. For a global assessment of distance accuracy, qualitatively (or quantitatively) examine the grid image and verify uniform grid pattern spacing without distortion.

8.4.9 Erasure Thoroughness

The ability to reuse the IP without residual signals from previous overexposures is important, and the erasure test evaluates the ability of the read/erase cycle to remove ghosting artifacts under severe exposure conditions as described in Table 11. If improperly or insufficiently erased, image artifacts superimposed on subsequent images can mimic disease processes. In the case of

Table 11. Testing and Acceptance Criteria for Erasure Thoroughness

	Agfa	Fuji	Kodak
Exposure condition	Place a lead block at the center of a 35×43 cm cassette and fully expose to ~50 mR with a 60 kVp beam (no added filtration) at 180 cm SID. Process the IP, and expose a second time to 1 mR without the lead object and the collimator positioned in 5 cm on each side of the IP. Process the IP, and then reprocess one more time using "dark noise" settings for the quantitative test. Three images total are used in this test.		
IP processing	System diagnosis flat field, speed class = 200	Test/sensitivity Semi EDR, first 2 images; for 3rd processing, use test/sensitivity (L = 1), fixed EDR (S = 10,000)	Pattern
Image post-processing	None Musica parameters = 0.0 Sensitometry = linear Window setting default or equivalent to 1 log(E) unit	Linear GA = 1.0, GT = A, RE = 0.0 Window setting default or equivalent to 1 log(E) unit	Raw data and no edge enhancement settings, Level = EI, Window setting defaultor equivalent to 1 log(E) unit
Measurements	lgM, avg pixel value (PV), and standard deviation (PVSD) and scan average level (SAL) within 80% of the 3rd image area.	Avg pixel value (PV) and standard deviation (PVSD) within 80% of the 3rd image area.	Exposure index (EI), avg pixel value (PV), and its standard deviation within 80% of the 3rd image area.
Qualitative criteria	Absence of a ghost image of the lead block in the 2nd image[a].		
Quantitative criteria (determined from analysis of 3rd image)	lgM = 0.28 SAL < 130 PV < 630 PVSD < 5	PV < 280[b] PVSD < 4	$EI_{GP} < 80$, $EI_{HR} < 380$ $PV_{GP} < 80$, $PV_{HR} < 80$ PVSD < 4

[a]In some systems the erasure time is configurable (e.g., Agfa).
[b]For an inverse relationship between PV and log(E), PV should be greater than 744.
[Adapted from Table XI in reference 56 with permission from AAPM.]

severe overexposure, an IP could possibly require several primary "erasures" and a rest interval to totally remove the residual signals.

This test is performed with high overexposure (open field) and underexposure (under a lead block) exposing an IP to an incident exposure of ~50 mR. After processing and cycling the IP, a second, clinically relevant incident exposure of ~1 mR is acquired without the lead block and with the collimator field positioned inward 5 cm on each side. The second exposed IP is processed using the normal processing cycle, followed by direct reinsertion into the reader and processing with the "dark noise" settings, for a total of three output images. Absence of a ghost image of the lead block object in the second image demonstrates acceptable erasure thoroughness. The third image should not contain digital image values above that specified in Table 11.

8.4.10 Aliasing/Grid Response

Moiré patterns are aliased signals that manifest as lower frequency "beat" patterns superimposed on the image, and are most often caused by the projection of the bar patterns of an anti-scatter grid. They are caused by high-frequency repeating patterns insufficiently sampled during the digitization process, occurring during the acquisition or during image manipulation and/or sub-sampling of the digital image matrix to fully fit the larger image matrix on a display that supports only a smaller matrix size. Stationary, low-frequency grids can readily produce an aliased pattern when the grid strips are aligned parallel to the scan or the plate translation direction; in fact, patterns are less readily manifested in the scan direction because of the application of "anti-aliasing" frequency filters applied to the scan line information. Aliasing artifacts caused by display sub-sampling can be distinguished by their changing appearance as the image size on the display changes. This is a consequence of inappropriate downsizing (instead of pixel averaging), where the high-frequency signals are maintained and aliased on the display.

The aliasing test should be performed for each type and size of IP in common use. Table 12 lists the exposure conditions for this test. Comparing the response with and without a grid allows the determination of potential grid problems. Moiré patterns should not be visible with grid lines perpendicular to scan direction (the preferred orientation of the grid), while grids of relatively low frequency (<50 lines/cm) will likely produce an aliasing pattern in the direction parallel to the scan direction. For moving grids, moiré patterns should not be visible in either direction. High-frequency stationary grids (e.g., >70 lines/cm) should have minimal aliasing (except perhaps for high-resolution systems dedicated to mammography applications).

8.4.11 IP Throughput

A feature that is particularly relevant to PSP system cost is the speed of IP processing and throughput. High-speed systems have external stackers or auto-load capabilities that allow the technologist to insert multiple cassettes and allow the system to scan and process the IPs automatically. This has an advantage in workflow as well as "pipelining" that allows the reader to extract the latent image and erase a previously scanned IP at the same time.

This test can be accomplished in conjunction with section 8.4.3. Table 13 describes the methodology and acceptable results based upon evaluation of a small number of cassettes and time needed to read and display the output image. The use of four cassettes slightly underestimates the IP throughput of the system in "stacker" readers because the first IP does not achieve the pipelining benefit of subsequent IPs. Pipelined processing realizes an increased throughput for several screens in the stacker, so the time for one processed image (display and film output)

Table 12. Testing and Acceptance Criteria for Aliasing/Grid Response

	Agfa	Fuji	Kodak
Exposure condition	Place the imaging plate/cassette in a bucky that contains an antiscatter grid so that the grid lines are *parallel* to the laser-scan direction. Alternatively, a grid may be placed directly on the screen. Make sure the grid movement is disabled. Expose the screen to 1 mR using an 80 kVp beam filtered with 0.5 mm Cu/1 mm Al filter and a SID according to the specification of the grid. Repeat, placing the screen perpendicular to the laser-scan direction. Repeat the exposure with a moving grid.		
IP Processing	System diagnosis flat field, speed class = 200	Test/contrast Semi EDR	Pattern
Image post-processing	None Musica parameters = 0.0 Sensitometry = linear Narrow window setting	Linear GA = 1.0, GT = A, RE = 0.0 Narrow window setting	Raw data and no edge enhancement settings Level = EI Narrow window setting
Measurements		None	
Qualitative criteria	Moiré pattern should not be visible with grid lines perpendicular to scan direction. For moving grids, no moiré pattern should be visible in either direction.		
Quantitative criteria	None		

[Adapted from Table XII in reference 56 with permission from AAPM.]

Table 13. Testing and Acceptance Criteria for IP Throughput

	Agfa	Fuji	Kodak
Exposure condition	Expose four IPs of the same dimension to 80 kVp and 2 mR using the standard geometry. Process the cassettes sequentially without delay. For completeness, test each IP size available.		
IP processing	System diagnosis flat field, speed class = 200	Test/contrast Semi EDR	Pattern
Image post-processing	Musica parameters typical of clinical usage	Irrelevant	None
Measurements	Time interval, t, in minutes between putting the first cassette in and display of the last image at the PSP workstation.[a] Throughput (IP/h) = 60 × 4 / t.		
Qualitative criteria		None	
Quantitative criteria	Throughput should be within 10% of the system specification for a given IP size.		

[a]The contribution of time delay caused by the network is not considered in this test.
[Adapted from Table XIII in reference 56 with permission from AAPM.]

will be longer than the average time for a series of screens. Throughput should be within 10% of the published specifications (unless other exceptions are made in the purchase agreement). If the test does not achieve the claimed throughput, a repeat of the procedure with 10 IPs will introduce minor error. In a clinical situation, the maximum throughput might not be achieved because of excessively overexposed plates (e.g., uncollimated regions in a high-exposure exam) in which the erasure stage of the PSP process can be the limiting step. This is particularly true for sequential, single slot PSP readers.

8.4.12 Acceptance Criteria and Quantitative Relationships

Table 14 describes unified acceptance criteria by expressing signal levels and standard deviations in terms of corresponding exposure levels.[56] Table 15 describes the relationship between incident exposure, the corresponding pixel value under specific processing conditions, and the incident exposure to exposure index value.

8.4.13 Image Processing: LUT Transforms and Frequency Enhancement

The intent is to verify the proper function of the algorithms provided by the manufacturers regarding specific image processing and user selected adjustments for clinical applications for contrast and frequency enhancement. Images of the low-contrast phantom, an aluminum step wedge, and high-resolution test phantom can demonstrate the effects of the image processing parameters and their impact on image quality as a first-pass test; however, there is no substitute for manipulation of clinical images.

Review of clinical images and collaboration with the radiologists to adjust image processing parameters to their liking is important. Vendor PSP imaging specialists should be available after completion of the equipment installation to assist in the optimization of exam algorithms on a case-by-case basis, and to train technologists, physicists, and radiologists in the operation of the PSP system, including minor image processing algorithm adjustments by the physicist and/or interested QC technologist. Default processing variables/parameters should be verified with published standard values for all examination codes. Unique situations requiring specific settings such as a grid versus no grid examination should be undertaken as well.

9. ARTIFACTS

9.1 Image Artifacts

Artifacts on images can originate from the hardware (e.g., x-ray system, grid, PSP reader, PSP detector), software (e.g., glitches, algorithms), or the object (e.g., positioning, motion, etc.).[99–103] Hardware artifacts arise from the image plate, the image reader, the hard-copy printer, or the processor. Most common are IP defects that are temporary and are likely due to dust, dirt, or phantom (non-erased) images. These artifacts can be easily corrected by screen cleaning and/or plate erasure. Permanent IP artifacts can be traced to scratches or screen aging—replacement will likely be necessary. The image reader can malfunction causing skipped scan lines and/or distorted images. Laser power will diminish over time to a point beyond correction (lifetime is estimated to be years, depending on daily use), necessitating replacement of the laser subsystem. Dust particles on the galvanometer/polygonal deflection mirror or on the light collection device

Table 14. PSP Response Tolerance Levels for "Uniform" Acceptance Criteria.
All signal levels and standard deviations are expressed in terms of corresponding exposure (E) values deduced from those quantities.

Characteristics	Quantity of Interest	Acceptable Tolerance
Dark noise	Average signal and its standard deviation within 80% of the image area	$E < 0.012$ mR $\sigma_E / E < 1\%$
Uniformity	Signal standard deviation within 80% of the image area, and the standard deviation of the average screen signal among screens	$\sigma_E < 5\%$
Exposure calibration	The exposure indicator response expressed in terms of exposure to 1 mR entrance exposure	$E_{measured} = 1 \pm 10\%$
Linearity and autoranging	The slope of the system response (expressed in terms of logarithm of exposure) vs. logarithm of actual exposure	Slope $= 1 \pm 10\%$ Correlation coefficient > 0.95
Laser beam function	Jitter dimension in pixels	Occasional jitters $< \pm 1$ pixel
Limiting resolution	Maximum discernible spatial frequencies of a high-contrast line-pair along scan, sub-scan, and 45° axes	$R_{scan} / f_{Nyquist} > 0.9$ $R_{sub\text{-}scan} / f_{Nyquist} > 0.9$ $R45° / (1.41 f_{Nyquist}) > 0.9$
Noise and low-contrast resolution	A linear fit of system noise (expressed in terms of logarithm of corresponding σ_E / E to logarithm of actual exposure	Correlation coefficient > 0.95
Spatial accuracy	The difference between the measured (d_m) and actual distances (d_0) in orthogonal directions	$(d_m - d_0)/ d_0 < 2\%$
Erasure thoroughness	Average signal and its standard deviation within 80% of the reread/unexposed image	$E < 0.012$ mR $\sigma_E / E < 1\%$
Aliasing/grid response	No quantitative tolerance levels	
IP throughput	Measured throughput in screens per hours (T_m) and the specified throughput (T_0)	$(T_0 - T_m) / T_0 < 10\%$

[Adapted from Table XIV in reference 56 with permission from AAPM.]

can be manifested as image dropout artifacts. Laser hard-copy printer misalignment and/or film conveyor malfunction can cause an uneven scan line distribution, image distortion, or shading, among a myriad of potential problems. Film processor artifacts should be considered as well.

9.2 Software Artifacts

Improper selection of processing menus resulting in incorrect histogram normalization, dynamic range scaling, and output film density are the major causes of software artifacts. The histogram analysis function may incorrectly identify the pixel values of interest in the image. Causes

Table 15. Relationship Between Exposure and Pixel Value/Exposure Indicator Responses.
PV is the pixel value, E is the exposure in mR, B is the speed class, and L is the latitude of the system, where appropriate.

	Agfa	Fuji	Kodak
Exposure indicator quantities	lgM and scan average level (SAL)	Sensitivity (S)	Exposure Index (EI)
Exposure indicator relationship	$SAL = 90\sqrt{(0.877cBE)}$ $lgM = 2 \log(SAL) - 3.9478$ $\quad = \log(cBE) - 0.0963$ $c = 1.0$ for MD30/40 IPs	$S = 200/E$	$EI = 1000 \log(E) + 2000$
Pixel value relationships	$PV = 2499 \log(SAL) - 4933$ $\quad = 1250 \log(cBE) - 121$[a] $c = 1.0$ for MD30/40 IPs	$PV = (1024/L)$ $\times (\log E + \log(S/200)$ $\quad + 511$	$PV = 1000 \log(E) + c_0$ $c_0 = 2000$ for GP IPs $c_0 = 1700$ for HR IPs
Exposure/ reading condition	75 kVp and 1.5 mm Cu filtration, no reading delay	80 kVp without filtration, 10 min reading delay	80 kVp and 0.5 mm Cu/ 1 mm Al, 15 min reading delay

[a]For a 12-bit, linear $\log(E)$ data transfer.
[Adapted from Table XIV in reference 56 with permission from AAPM.]

include mispositioning of the object, collimation detection errors that can occur in high scatter situations, and unusual anatomic variations that confuse the algorithms that identify the useful image information on the detector.

9.3 Object Artifacts

These artifacts usually arise due to object mispositioning as described above, scan line interference patterns with the grid resulting in obvious moiré patterns, cassettes positioned upside down, random dropouts, or high pass frequency processing. If not properly adjusted, a "halo" effect could appear around the edges of objects by the unsharp masking technique. Backscatter can contribute significantly to contrast degradation when a substantial scattering volume is behind the cassette, possibly introducing phantom images such as the "tombstone" artifact.[16]

9.4 Film Artifacts

Fogging, pressure marks, static electricity discharge, improper processing caused by inadequate or contaminated chemistry or inappropriate temperature levels of the developer/fixer, putting the film in upside down in the laser printer, and other similar errors will result in the manifestation of artifacts attributed to film.

10. QUALITY CONTROL AND PERIODIC MAINTENANCE

Periodic QC testing is necessary for checking system performance and maintaining optimum image quality. Daily, monthly, quarterly, and annual procedures are recommended as part of an ongoing QC program. In most cases, except for major problems and yearly tests, an assigned technologist can perform most of the tasks. A QC phantom specifically designed for PSP radiography should be purchased with the system (they are often optional items). In addition, automated QC methods for system checks, monitor maintenance/setup, and adjustments should be requested from the manufacturers. Database management tools and spreadsheets are very powerful quantitative and graphical analyzers of pertinent system performance, and can assist in the identification of problem areas. The following is a suggested QC and performance evaluation schedule that may be used as a guideline, in addition to any routine QC activities and preventive maintenance program recommended by the manufacturer of the individual PSP system. Depending on the specific PSP system characteristics and resources available, some tests may be unnecessary, or frequencies may be adjusted.

10.1 Daily Tests (Technologist)

1. Inspect system operation and verify operational status.

2. Create laser-generated sensitometry strip and measure film densities (if applicable).

3. Erase cassettes before use, if unsure of status.

4. When performing image QC, look for dust particles, scratches, and mechanical friction marks in the images. If present, immediately remove the corresponding cassette/IP and clean as necessary according to the manufacturer's instructions. Remove IPs that cannot adequately be cleaned from the inventory. Ensure all images are QC'ed and verified. Check status of network queue and send images as necessary to PACS.

10.2 Monthly Tests (Technologist)

1. Erase all plates in the inventory (particularly those that are infrequently used), and perform spot checks on randomly chosen plates, testing according to methods described in Table 3 to ensure low Dark Noise.

2. Acquire QC phantom image and implement QC measurements (vendor specific, hopefully automated). Verify performance levels; store results in database file. *Note:* This QC evaluation might be useful on a more frequent (weekly) basis, depending on ease of use or as recommended by the PSP manufacturer for routine QC.

3. Verify calibration of PSP QC workstations, using a simple qualitative SMPTE video test pattern[85] and/or the methodology of AAPM TG-18 for monitor quality control.[70]

10.3 Quarterly Tests (Technologist)

1. A cleaning program for all cassettes and imaging plates is necessary; inspect, clean the imaging plates with approved cleaner by manufacturer, erase and put back into inventory. *Note:* The cleaning frequency depends on the specific conditions at the

site, and can be as frequent as *once per week* in particular environments, or as infrequent as once per year.

2. Perform qualitative and quantitative QC phantom analysis, including resolution, contrast/noise, laser jitter, and exposure indicator accuracy.

3. Review image retake rate; determine causes of unacceptable PSP images.

4. Review QC exposure indicator database; determine cause of under/overexposures, implement corrective action; generate quarterly reports.

10.4 Annually and After Major Repair/Calibration (Physicist)

1. Inspection/evaluation of image quality; spot check image processing algorithms for appropriateness.

2. Acceptance test procedures to verify and/or re-establish baseline values. Use complete inspection and verification procedures.

3. Review technologist QC activities and reports; evaluate retake activity, patient exposure trends, QC records, and service history of the equipment.

In addition to the periodic testing, all inspections should be done on an "as needed" basis, particularly for hardware/software changes and significant repairs/changes to the equipment. The designated QC technologist, physicist, and service personnel should participate in the preventive maintenance and QC program. Invasive adjustments or corrections of the PSP system should be done only with "vendor-approved" personnel, and with the knowledge of the technologists, physicist, and other service personnel responsible for quality control.

11. CONCLUSIONS

Computed radiography using PSP detectors is becoming more widespread and clinically important, as the replacement of screen-film detectors continues. The "cassette-based" x-ray acquisition and general radiographic technique methods are in many ways similar to screen-film; however the considerable differences must also be recognized. A large number of setup parameters, inappropriate use of exposure menus, system hardware or software malfunctions, plate damage, excessive quantum mottle, and patient positioning details are among a host of potential issues that differ from screen-film experiences. Technologist and radiologist training must address the unique attributes and image acquisition rules (collimation, autoranging, image processing), as well as continuing education of PSP radiography. As proliferation of systems and manufacturers occur, unique characteristics, controls, and testing procedures must be considered, in addition to the "generic" operational aspects of PSP and PSL. The acceptance test and QC procedures of a PSP system are reasonably straightforward and relatively easy to evaluate. Widespread use of such systems is occurring rapidly and the technology is evolving. Acceptance test and QC procedures must advance as well to ensure and maintain optimal image quality. This report provides a reasonable starting point.

APPENDIX A. MANUFACTURER CONTACT INFORMATION

Manufacturers of PSP system products; web addresses (not inclusive; as of October 2005). Each of the vendors has specific protocols developed for their systems; in many cases, these are patterned after the AAPM Task Group 10 documentation. In the early stages of document formulation, these protocols were part of the appendices, but due to the rapid change in technology and equipment, were removed. This information should be directly obtainable from the vendors by requesting the information regarding acceptance testing and quality control of the PSP equipment.

- FujiFilm Medical Systems USA, Inc.
  http://www.fujimed.com/

- Eastman Kodak
  http://www.kodak.com/global/en/health/productsByType/cr/crProduct.jhtml

- Agfa
  http://www.agfa.com/healthcare

- Konica
  http://konicaminolta.us

- Orex (purchased by Kodak)
  http://www.orex-cr.com

- iCRco (Cobrascan)
  http://www.icrcompany.com

- ALARA, Inc.
  http://www.alara.com

- Other manufacturer original equipment manufacturer branding of PSP system products
  – Philips Medical Systems
  – Siemens Medical Systems
  – General Electric Healthcare
  – Toshiba Medical Systems.

APPENDIX B. AVAILABLE ACCEPTANCE TESTING AND QUALITY CONTROL FORMS

Ehsan Samei Microsoft® Excel® Spreadsheets

Downloadable spreadsheets that provide objective criteria for many of the Task Group 10 recommended tests are available at: http://deckard.mc.duke.edu~samei/files, for specific testing of Fuji, Agfa, and Kodak PSP systems. It is prudent to consult the manufacturer for the latest specifications of the equipment.

REFERENCES

1. Digital Imaging and Communications in Medicine (DICOM). http://www.nema.org. Rosslyn, VA: National Electrical Manufacturers Association (NEMA). (2004).

2. Health Level 7 (HL7) Standard http://www.hl7.org. Health Level 7, Inc., 3300 Washtenaw Ave., Suite 227, Ann Arbor, MI 48104. (2004)

3. J. A. Rowlands. "The physics of computed radiography." *Phys Med Biol* 47:123–166 (2002).

4. K. P. Andriole. "Productivity and cost assessment of computed radiography, digital radiography, and screen-film for outpatient chest examinations." *J Digit Imaging* 15:161–169 (2002).

5. K. P. Andriole, D. M. Luth, and R. G. Gould. "Workflow assessment of digital versus computed radiography and screen-film in the outpatient environment." *J Digit Imaging* 15 (Suppl 1):124–126 (2002).

6. D. M. Deleeuw, T. Kovats, and S. P. Herko. "Kinetics of photostimulated luminescence in BaFBr-Eu." *JES. J Electrochem Soc* 134:491–493 (1987).

7. K. Takahashi, Y. Shibahara, and J. Miyahara. "Photostimulated luminescence (PSL) and color-centers in BaFCl-Eu2+, BaFBr-Eu2+, BaFI-Eu2+ phosphors." *JES J Electrochem Soc* 130:C245–C245 (1983).

8. K. Takahashi, K. Kohda, J. Miyahara, Y. Kanemitsu, K. Amitani, and S. Shionoya. "Mechanism of photostimulated luminescence in BaFCl-Eu-2+, BaFBr-Eu-2+ phosphors." *J Lumin* 31(2):266-268 (1984).

9. H. von Seggern, T. Voight, W. Knupfer, and G. Lange. "Physical model of photostimulated luminescence of x-ray irradiated BaFBr:Eu." *J Appl Phys* 64:1405–1412 (1988).

10. A. J. Wilson and O. C. West. "Single exposure conventional and computed radiography. The hybrid cassette revisited." *Invest Radiol* 28:409–412 (1993).

11. H. MacMahon, S. Sanada, K. Doi, M. Giger, X. W. Xu, F. F. Yin, S. M. Montner, and M. Carlin. "Direct comparison of conventional and computed radiography with a dual-image recording technique." *Radiographics* 11:259–268 (1991).

12. G. Luckey. Apparatus and Methods for Producing Images Corresponding to Patterns of High Energy Radiation. U. S. Patent No. 3859527. (1975).

13. S. Arakawa, W. Itoh, K. Kohda, and T. Suzuki. "Novel computed radiography system with improved image quality by detection of emissions from both sides of an imaging plate." *Proc SPIE* 3659:572–581 (1999).

14. P. J. Leblans, L. Struye, and P. Willens. "New needle crystalline CR detector." *Proc SPIE* 4320:59–67 (2001).

15. R. Schaetzing. "Management of pediatric radiation dose using Agfa computed radiography." *Pediatr Radiol* 34(Suppl 3):S207–S214 (2004).

16. D. M. Tucker, M. Souto, and G. T. Barnes. "Scatter in computed radiography." *Radiology* 188:271–274 (1993).

17. H. Kato. "Photostimulable Phosphor Radiography: Design Considerations" in *Specification, Acceptance Testing and Quality Control of Diagnostic X-ray Imaging Equipment,* edited by J. A. Seibert, G. T. Barnes, and R. G. Gould. AAPM Medical Physics Monograph No. 20. Woodbury, NY: American Institute of Physics, Inc., pp. 731–770, 1994.

18. S. Arakawa, H. Yasuda, K. Kohda, and T. Suzuki. "Improvement of image quality in CR mammography by detection of emissions from dual sides of an imaging plate." *Proc SPIE* 3977:590–600 (2000).

19. T. Bogucki, D. Trauernicht, and T. Kocher. Characteristics of a Storage Phosphor System for Medical Imaging. Kodak Health Sciences Division. Rochester, NY: Eastman Kodak Company (1995).

20. J. A. Seibert, J. M. Boone III, V. N. Cooper, and K. K. Lindfors. "Cassette-based digital mammography." *Technol Cancer Res Treat* 3:413–427 (2004).

21. K. A. Fetterly and B. A. Schueler. "Performance evaluation of a 'dual-side read' dedicated mammography computed radiography system." *Med Phys* 30:1843–1854 (2003).

22. R. Schaetzing, R. Fasbender, and P. Kersten. "New high-speed scanning technique for computed radiography." *Proc SPIE* 4682:511–520 (2002).

23. S. Arakawa, T. Hiroaki, H. Kuwabara, T. Suzuki, and T. Hagiwara. "Compact high-speed computed radiography (CR) system using a linear CCD with a large-area photodiode (PD) and dual transfer lines." *Proc SPIE* 5030:778–787 (2004).

24. M. J. Flynn. "Processing Digital Radiographs of Specific Body Parts" in A*dvances in Digital Radiography: Categorical Course in Diagnostic Radiology Physics,* edited by E. Samei and M. Flynn. Oak Brook, IL: Radiological Society of North America (RSNA), 2003.

25. J. A. Seibert. "Digital Image Processing Basics" in *A Categorical Course in Physics: Physical and Technical Aspects of Angiography and Interventional Radiology,* edited by S. Balter and T. B. Shope. Oak Brook, IL: Radiological Society of North America (RSNA), pp. 123–142, 1995.

26. J. A. Seibert. "Digital Radiographic Image Presentation: Preprocessing Methods" in *Advances in Digital Radiography: Categorical Course in Diagnostic Radiology Physics,* edited by E. Samei and M. J. Flynn. Oak Brook, IL: Radiological Society of North America (RSNA), pp. 63–70, 2003.

27. J. A. Seibert, D. K. Shelton, and E. H. Moore. "Computed radiography x-ray exposure trends." *Acad Radiol* 3:313–318 (1996).

28. P. Vuylsteke, P. Dewaele, and E. Schoeters. "Optimizing Computed Radiography Imaging Performance" in *The Expanding Role of Medical Physics in Diagnostic Imaging,* edited by G. D. Frey and P. Sprawls. AAPM Medical Physics Monograph No. 23. Madison, WI: Advanced Medical Publishing, pp. 107–151, 1997.

29. "Automatic Setting Functions for Image Density and Range in the FCR System." *Computed Radiography, Technical Review No. 3.* Tokyo: Fuji Photo Film Co. Ltd. (1993).

30. M. Ogoda, K. Hishinuma, M. Yamada, and K. Shimura. "Unsharp masking technique using multiresolution analysis for computed radiography image enhancement." *J Digit Imaging* 10:185–189 (1997).

31. P. Vuylsteke and E. Schoeters. "Multiscale image contrast amplification (MUSICA)." *Proc SPIE* 2167:551–560 (1994).

32. R. Van Metter and D. Foos. "Enhanced latitude for digital projection radiography." *Proc SPIE* 3658:468–483 (1999).

33. A. W. Templeton, L. H. Wetzel, L. T. Cook, L. A. Harrison, D. A. Eckard, W. H. Anderson, and K. S. Hensley. "Enhancement of storage phosphor plate images: A C-language program." *J Digit Imaging* 5:59–63 (1992).

34. "Fuji Technical Review on Image Processing." *Computed Radiography, Technical Review No. 3.* Tokyo: Fuji Photo Film Co. Ltd. (1993).

35. R. E. Alvarez, J. A. Seibert, and S. K. Thompson. "Comparison of dual energy detector system performance." *Med Phys* 31:556–565 (2004).

36. D. L. Ergun, C. A. Mistretta, D. E. Brown, R. T. Bystrianyk, W. K. Sze, F. Kelcz, and D. P. Naidich. "Single-exposure dual-energy computed radiography: Improved detection and processing." *Radiology* 174:243–249 (1990).

37. W. Ito, K. Shimura, N. Nakajima, M. Ishida, and H. Kato. "Improvement of detection in computed radiography by new single-exposure dual-energy subtraction." *J Digit Imaging* 6:42–47 (1993).

38. F. Kelcz, F. E. Zink, W. W. Peppler, D. G. Kruger, D. L. Ergun, and C. A. Mistretta. "Conventional chest radiography vs dual-energy computed radiography in the detection and characterization of pulmonary nodules." *AJR Am J Roentgenol* 162:271–278 (1994).

39. S. Kido, H. Nakamura, W. Ito, K. Shimura, and H. Kato. "Computerized detection of pulmonary nodules by single-exposure dual-energy computed radiography of the chest (part 1)." *Eur J Radiol* 44:198–204 (2002).

40. S. Kido, K. Kuriyama, C. Kuroda, H. Nakamura, W. Ito, K. Shimura, and H. Kato. "Detection of simulated pulmonary nodules by single-exposure dual-energy computed radiography of the chest: effect of a computer-aided diagnosis system (part 2)." *Eur J Radiol* 44:205–209 (2002).

41. B. K. Stewart and H. K. Huang. "Single-exposure dual-energy computed radiography." *Med Phys* 17:866–875 (1990).

42. A. Jonsson, K. Jonsson, K. Eklund, G. Holje, and H. Pettersson. "Computed radiography in scoliosis. Diagnostic information and radiation dose." *Acta Radiol* 36:429–433 (1995).

43. J. A. Kalmar, J. P. Jones, and C. R. Merritt. "Low-dose radiography of scoliosis in children. A comparison of methods." *Spine* 19:818–823 (1994).

44. K. P. Andriole, R. G. Gould, and W. R. Webb. "Finding-specific display presets for computed radiography soft-copy reading." *J Digit Imaging* 12:3–5 (1999).

45. E. L. Siegel and B. I. Reiner. "Challenges associated with the incorporation of digital radiography into a picture archival and communication system." *J Digit Imaging* 12:6–8 (1999).

46. J. Xuan, T. Adali, Y. Wang, and E. Siegel. "Automatic detection of foreign objects in computed radiography." *J Biomed Opt* 5:425–431 (2000).

47. W. Huda, L. N. Rill, and A. P. Bruner. "Relative speeds of Kodak computed radiography phosphors and screen-film systems." *Med Phys* 24:1621–1628 (1997).

48. L. W. Goldman. "Speed Values, AEC Performance Evaluation, and Quality Control" in *Specifications, Performance Evaluation, and Quality Assurance of Radiographic and Fluoroscopic Systems in the Digital Era,* edited by L. W. Goldman and M. V. Yester. Madison, WI: Medical Physics Publishing, pp. 271–297, 2004.

49. C. E. Willis. "Strategies for dose reduction in ordinary radiographic examinations using CR and DR." *Pediatr Radiol* 34:S196-S200 (2004).

50. S. J. Shepard and et al. "Standardization of Exposure Indices for Digital Radiography; AAPM Task Group 116." DR Exposure Index Task Group report in progress. (2005).

51. W. Huda, R. M. Slone, C. J. Belden, J. L. Williams, W. A. Cumming, and C. K. Palmer. "Mottle on computed radiographs of the chest in pediatric patients." *Radiology* 199:249–252 (1996).

52. J. T. Dobbins III, D. L. Ergun, L. Rutz, D. A. Hinshaw, H. Blume, and D. C. Clark. "DQE(f) of four generations of computed radiography acquisition devices." *Med Phys* 22:1581–1593 (1995).

53. G. T. Barnes. "Digital X Ray Image Capture with Image Intensifier and Storage Phosphor Plates: Imaging Principles, Performance and Limitations" in *Digital Imaging,* edited by J. Trueblood and W. Hendee. Proceedings of the 1993 AAPM Summer School, Medical Physics Monograph No. 22. Madison, WI: Medical Physics Publishing, pp. 23–48, 1993.

54. A. R. Lubinsky, B. R. Whiting, and J. F. Owen. "Storage phosphor system for computed radiography: Optical effects and detective quantum efficiency (DQE)." *Proc SPIE* 767:167–177 (1987).

55. W. Hillen, U. Schiebel, and T. Zaengel. "Imaging performance of a digital storage phosphor system." *Med Phys* 14:744–751 (1987).

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

57. J. M. Boone and J. A. Seibert. "An accurate method for computer-generating tungsten anode x-ray spectra from 30 to 140 kV." *Med Phys* 24:1661–1670 (1997).

58. Digital Imaging and Communications in Medicine (DICOM) Part 14: Grayscale Standard Display Function. http://medical.nema.org/dicom/2003/03_14PU.pdf. Rosslyn, VA: National Electrical Manufacturers Association (NEMA) (2003).

59. E. Samei et al. Assessment of Display Performance for Medical Imaging Systems: AAPM Task Group 18 report. Draft (2004).

60. P. C. Bunch. "Performance characteristics of high-MTF screen film systems." *Proc SPIE* 2163:14–35 (1994).

61. D. Clunie. "DICOM Softcopy Presentation State Storage and Print Presentation LUT." http://www.dclunie.com/papers/spie_mi_nema_softcopy.pdf. (2005).

62. D. S. Channin. "M:I-2 and IHE: Integrating the healthcare enterprise, year 2." *Radiographics* 20:1261–1262 (2000).

63. D. S. Channin, E. L. Siegel, C. Carr, and J. Sensmeier. "Integrating the healthcare enterprise: A primer. Part 5. The future of IHE." *Radiographics* 21:1605–1608 (2001).

64. D. S. Channin, C. Parisot, V. Wanchoo, A. Leontiev, and E. L. Siegel. "Integrating the healthcare enterprise: A primer: Part 3. What does IHE do for ME?" *Radiographics* 21:1351–1358 (2001).

65. D. S. Channin. "Integrating the healthcare enterprise: A primer. Part 2. Seven brides for seven brothers: The IHE integration profiles." *Radiographics* 21:1343–1350 (2001).

66. D. S. Channin. "Integrating the healthcare enterprise: A primer. Part 6: The fellowship of IHE: Year 4 additions and extensions." *Radiographics* 22:1555–1560 (2002).

67. M. Henderson, F. M. Behlen, C. Parisot, E. L. Siegel, and D. S. Channin. "Integrating the healthcare enterprise: A primer. Part 4. The role of existing standards in IHE." *Radiographics* 21:1597–1603 (2001).

68. S. M. Moore. "Using the IHE scheduled work flow integration profile to drive modality efficiency." *Radiographics* 23:523-529 (2003).

69. M. Oskin, J. A. Seibert, and R. L. Kennedy. "Diagnostics and Acceptance Testing for DICOM Networks" in *Practical Digital Imaging and PACS,* edited by J. A. Seibert, L. J. Filipow, and K. P. Andriole. AAPM Medical Physics Monograph No. 25. Madison, WI: Medical Physics Publishing, pp. 317–350, 1999.

70. E. Samei, A. Badano, D. Chakraborty, K. Compton, C. Cornelius, K. Corrigan, M. J. Flynn, B. Hemminger, N. Hangiandreou, J. Johnson, D. M. Moxley-Stevens, W. Pavlicek, H. Roehrig, L. Rutz, J.

Shepard, R. A. Uzenoff, J. Wang, and C. E. Willis. "Assessment of display performance for medical imaging systems: Executive summary of AAPM TG18 report." *Med Phys* 32:1205–1225 (2005). Full report available as AAPM On-line Report No. O3 (2005).

71. H. P. Chan and K. Doi. "Investigation of the performance of antiscatter grids: Monte Carlo simulation studies." *Phys Med Biol* 27:785–803 (1982).

72. D. Gelskey and R. LaFrance. "RadChek: Electronic device to calibrate AEC phototimer for CR." Unpublished. (2004).

73. E. G. Christodoulou, M. M. Goodsitt, H. P. Chan, and T. W. Hepburn. "Phototimer setup for CR imaging." *Med Phys* 27:2652–2658 (2000).

74. L. J. Cesar. "Computed radiography: Its impact on radiographers." *Radiol Technol* 68:225–232 (1997).

75. D. S. Artz. "Computed radiography for the radiological technologist." *Semin Roentgenol* 32:12–24 (1997).

76. A. Workman, A. R. Cowen, and D. S. Brettle. "Physical evaluation of computed radiography as a mammographic x-ray imaging system." *Br J Radiol* 67:988–996 (1994).

77. C. E. Willis, J. Weiser, R. G. Leckie, J. R. Romlein, and G. Norton. "Optimization and quality control of computed radiography." *Proc SPIE* 2164:178–185 (1994).

78. J. E. Tucker, M. Contreras, R. J. Wider, M. G. Radvany, A. K. Chacko, and R. B. Shah. "Photostimulable storage phosphor image acquisition: Evaluation of three commercially available state-of-the-art systems." *J Digit Imaging* 12:54–58 (1999).

79. J. A. Seibert. "Photostimulable Phosphor Acceptance Testing" in *Specification, Acceptance Testing and Quality Control of Diagnostic X-ray Imaging Systems,* edited by J. A. Seibert, G. T. Barnes, and R. G. Gould. AAPM Medical Physics Monograph No. 20. Woodbury, NY: American Institute of Physics, Inc., pp. 771–800, 1994.

80. S. M. Kengyelics, J. H. Launders, and A. R. Cowen. "Physical imaging performance of a compact computed radiography acquisition device." *Med Phys* 25:354–360 (1998).

81. H. Jafroudi, D. Steller, M. Freedman, and S. K. Mun. "Quality control of storage phosphor digital radiography systems." *Proc SPIE* 2432:563–577 (1995).

82. W. Huda, M. Arreola, and Z. Jing. "Computed radiography acceptance testing." *Proc SPIE* 2432:512–521 (1995).

83. A. R. Cowen, A. Workman, and J. S. Price. "Physical aspects of photostimulable phosphor computed radiography." *Br J Radiol* 66:332–345 (1993).

84. E. Samei. Excel Spreadsheets for Acceptance Testing Criteria. http://deckard.mc.duke.edu/~samei/files. (2001).

85. J. E. Gray. "Use of the SMPTE test pattern in picture archiving and communication systems." *J Digit Imaging* 5:54–58 (1992).

86. A. Haus and S. Jaskulski. *The Basics of Film Processing in Medical Imaging.* Madison, WI: Medical Physics Publishing, 1997.

87. S. J. Shepard, P. P. Lin et al. AAPM Report No. 74: Quality Control in Diagnostic Radiology. Report of Task Group 12, Diagnostic X-ray Imaging Committee. Madison, WI: Medical Physics Publishing (2002).

88. L. K. Wagner and S. Dinwiddie. "Acceptance Testing and QC of Film Transport and Processing Systems" in *Specification, Acceptance Testing and Quality Control of Diagnostic X-ray Imaging Equipment,* edited by J. A. Seibert, G. T. Barnes, and R. G. Gould. AAPM Medical Physics Monograph No. 20. Woodbury, NY: American Institute of Physics, Inc., pp. 413–428, 1994.

89. J. A. Seibert. "Film Recording Printers" in *Quality Assurance: Meeting the Challenge in the Digital Medical Enterprise,* edited by B. I. Reiner, E. L. Siegel, and J. A. Carrino. Great Falls, VA: Society for Computer Applications in Radiology, pp. 93–110, 2002.

90. S. J. Shepard, S. K. Thompson, K. T. Krugh, S. Venable, and R. Munden. "Design and implementation of a robust high availability DICOM print network in a large multimodality clinical environment." *Proc SPIE* 4685:263–269 (2002).

91. S. K. Thompson, C. E. Willis, K. T. Krugh, S. J. Shepard, and K. W. McEnery. "Implementing the DICOM Grayscale Standard Display Function for mixed hard- and soft-copy operations." *J Digit Imaging* 15 (Suppl 1):27–32 (2002).

92. S. J. Shepard. "DICOM Basics for Radiographic and Fluoroscopic Systems" in *Specifications, Performance Evaluation, and Quality Assurance of Radiographic and Fluoroscopic Systems in the Digital Era,* edited by L. W. Goldman and M. V. Yester. AAPM Medical Physics Monograph No. 30. Madison, WI: Medical Physics Publishing, pp. 125–141, 2004.

93.   D. Clunie. "Designing and implementing a PACS-aware DICOM image object for digital x-ray, mammography, and intraoral applications." *Proc SPIE* 3662:83–89 (1999).

94.   J. McArdle. *Leeds X-ray Test Objects for Computed Radiography*. Instruction Manual. Leeds, England: FAXIL: Medical Physics (1990).

95.   A. J. Wagner, G. T. Barnes, and X. Z. Wu. "Assessing fluoroscopic contrast resolution: A practical and quantitative test tool." *Med Phys* 18:894–899 (1991).

96.   E. Samei, M. J. Flynn, and D. A. Reimann. "A method for measuring the presampled MTF of digital radiographic systems using an edge test device." *Med Phys* 25:102–113 (1998).

97.   J. Seibert and L. Goldman. "DR QC workshop." Digital Radiography QC Workshop at AAPM Annual Meeting, Seattle, WA, July 2005. *Med Phys* 32(6):2141 (2005).

98.   E. Samei and M. J. Flynn. "An experimental comparison of detector performance for computed radiography systems." *Med Phys* 29:447–459 (2002).

99.   M. J. Flynn and E. Samei. "Experimental comparison of noise and resolution for 2k and 4k storage phosphor radiography systems." *Med Phys* 26:1612–1623 (1999).

100.   L. J. Cesar, B. A. Schueler, F. E. Zink, T. R. Daly, J. P. Taubel, and L. L. Jorgenson. "Artefacts found in computed radiography." *Br J Radiol* 74:195–202 (2001).

101.   J. W. Oestmann, M. Prokop, C. M. Schaefer, and M. Galanski. "Hardware and software artifacts in storage phosphor radiography." *Radiographics* 11:795–805 (1991).

102.   S. L. Solomon, R. G. Jost, H. S. Glazer, S. S. Sagel, D. J. Anderson, and P. L. Molina. "Artifacts in computed radiography." *AJR Am J Roentgenol* 157:181–185 (1991).

103.   C. E. Willis, S. K. Thompson, and S. J. Shepard. "Artifacts and misadventures in digital radiology." *Appl Radiol* 33(1):11–17 (2004).