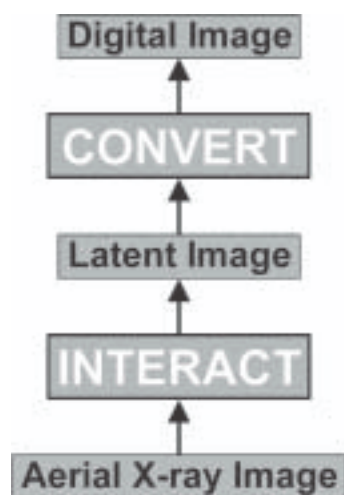


# Computed Radiography Technology<sup>1</sup>

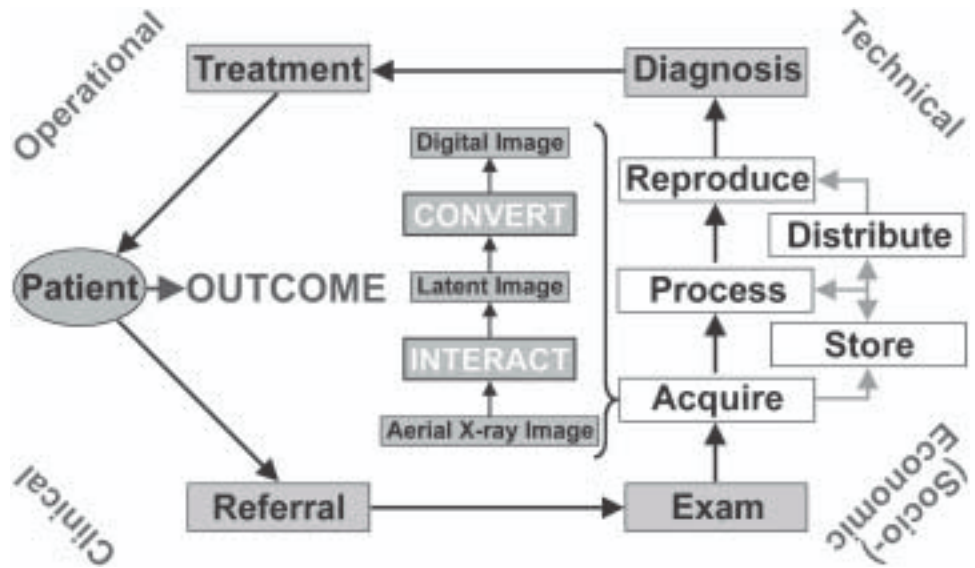
Although people often make a distinction between the technology that today is commonly called computed radiography (CR) and the family of projection imaging technologies called digital radiography (DR), CR is really a form of DR, in fact, its earliest form. To add to the confusion, CR itself also has a number of different names. In the radiology literature, the technology is referred to, among other things, as digital radiography with storage phosphors (SP), digital luminescence radiography (DLR), photostimulable luminescence (PSL) radiography, and variations on these terms, along with their abbreviations. Regardless of the name used (and several of them are deliberately used in this chapter), these names all refer to a technology designed to acquire or record projection images made with high-energy electromagnetic radiation, for example, x rays, on a reusable detector containing special storage phosphor materials (1–3).

Any image acquisition technology for digital projection radiography must do three things well (Fig 1): (a) interact with (eg, absorb) the analog x-ray aerial image emerging from an exposed object, usually a patient; (b) produce, and retain sufficiently long, a latent image corresponding to the aerial image; and (c) convert the latent image into a digital image. This acquisition process is only a small but important piece of the imaging system, which, in turn, is embedded in a much bigger chain of events (Fig 2) (1).



**Figure 1.** Chain of events in digital image acquisition. The aerial (x-ray) image emerging from an exposed object (patient) interacts with a radiation detector, which produces a latent image that is retained long enough to be measured and converted into a digital image.

**Figure 2.** Context of digital image acquisition. The image acquisition section is part of a larger imaging system that can also process, store, distribute, and reproduce images. The imaging system, in turn, is part of a diagnostic imaging chain that includes the diagnosis, treatment, and outcome for the patient. Finally, this diagnostic chain is embedded in a clinical, technical, operational, and socioeconomic context that influences each of the elements inside.



First, the acquired digital image must be processed to produce a new digital image suitable for human viewing. This optimized image must also be reproduced on some analog output medium or device (eg. film from a laser printer, soft-copy display) with which a human being can interact. In the case of machine vision (eg. computer-aided detection or diagnosis [CAD, CAD<sub>x</sub>]), the image processing extracts useful diagnostic information directly from the digital image data without necessarily producing an image as output. There must also be a means to manage the image data, for example, to store them and distribute them among the various components that make up the complete digital imaging system or the PACS (picture archiving and communication system).

The extended diagnostic imaging chain also includes a number of other important elements. The ultimate outcome for a patient referred for an imaging procedure depends only in part on the quality of the imaging system. The outcome depends even more on the quality of the diagnosis and on the resultant management or treatment decisions. Finally, this diagnostic chain is embedded in a clinical, technical, operational, and socioeconomic context that strongly influences its internal work flow and data flow. Thus, although the focus of this chapter is on digital image acquisition with a specific technology, one must always keep in mind the bigger picture that determines the ultimate use and usefulness of the acquisition system.

### TAXONOMY OF DIGITAL RADIOGRAPHY TECHNOLOGIES

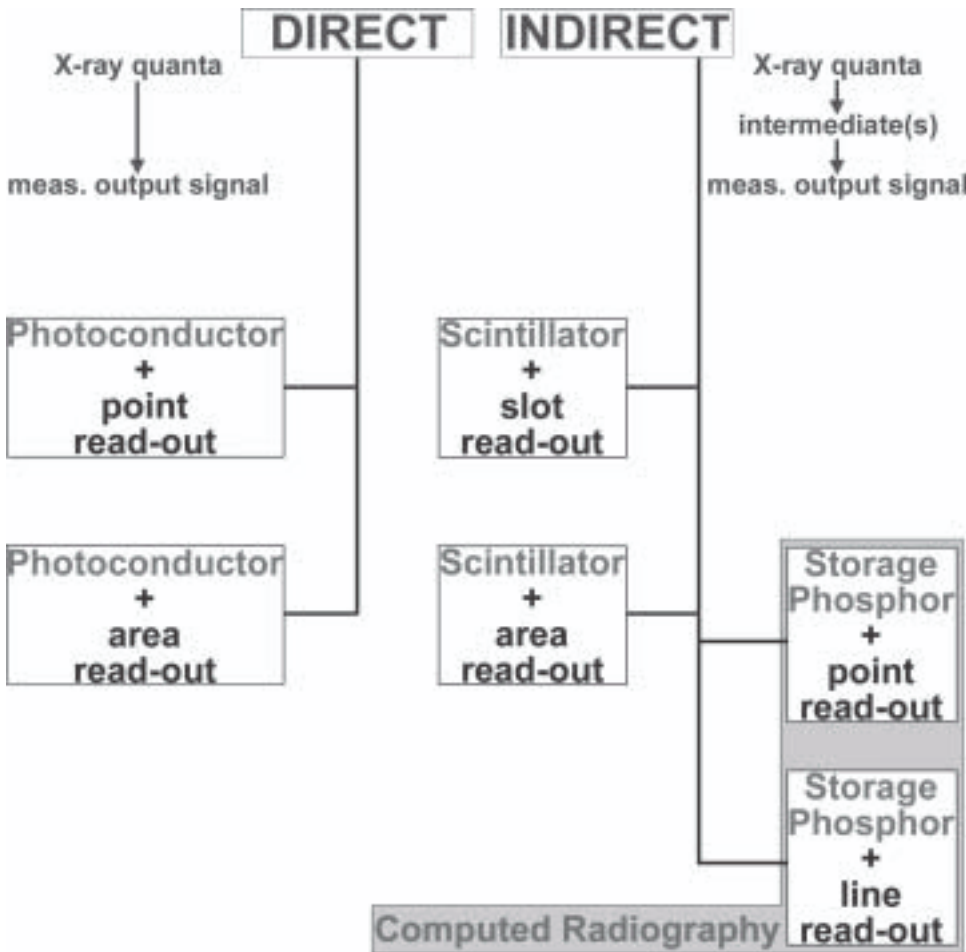
Some would argue that as long as the imaging system delivers images of sufficient quality for the customer's application, that is, as long as the imaging system is not the limiting step, then it is irrelevant which technology is

actually "under the hood." On the other hand, because the image acquisition technology creates the fiducial image that all subsequent links in the imaging chain must handle, this technology is the ultimate determinant of image quality in the objective, measurable sense, assuming that the actual imaging procedure was done properly. For this reason, some level of familiarity and comfort with the characteristics of the available technologies is certainly advisable. However, the variety of technologies and products for digital projection radiography is increasing every year, so it can sometimes be confusing to distinguish one type of technology from another.

Figure 3 shows a simple taxonomy of DR technologies based on (a) the number of conversion steps between the absorption of the incident x-ray quanta and the creation of a measurable latent-image signal, (b) the kind of x-ray detector used, and (c) the geometry of the latent-image readout process. Other chapters of this syllabus will address the details of the various DR technologies (4–6) in Figure 3, so the following descriptions will be brief.

As their name implies, direct DR technologies convert the incident x-ray quanta directly into a measurable latent-image signal. One early example of this type of DR (Philips Thoravision system) uses a uniformly charged photoconductive drum as the x-ray absorber. The incident x-ray aerial image discharges the drum imagewise, producing a static charge distribution that can be measured point by point with tiny electrometers riding just above the surface of the rotating drum. Another direct DR technology uses a photoconductor plate coupled to an active-matrix thin-film-transistor (TFT) array that reads the latent charge image in an area mode. This technology is used by several manufacturers (eg. Hologic DirectRay detector).

Indirect DR technologies convert the incident x-ray quanta into some intermediate state or states (eg. light



**Figure 3.** Taxonomy of DR technologies. Direct DR technologies convert the incident x-ray quanta directly into a measurable latent-image signal, while indirect DR has one or more intermediate states between these end points (see text for descriptions of technology). *meas* = measurable.

photons) before producing the measurable latent-image signal. An x-ray scintillator, for example, will absorb x rays and produce light that can be detected with a photodetector, which then produces an electric-charge latent image that can be measured. One example of an indirect DR technology (Fischer SenoScan TrueView) uses a slot-shaped x-ray beam with a slot-shaped x-ray detector and charge-coupled device (CCD) photodetector assembly that is scanned across the patient in synchrony with the beam. Area or full-field detectors that use a scintillator panel coupled to an active-matrix TFT array (eg, GE Revolution, Siemens AXIOM Aristos, Philips Digital Diagnost, Canon CXDI series) are also available, as are area detectors that use scintillators and area CCDs (eg, SwissRay dDR).

The scintillators used in indirect DR technologies emit light promptly on exposure to x rays. Another type of indirect DR, the one treated in this chapter, uses an x-ray detector designed to store a portion of the incident x-ray energy for later readout (1–3). In such a storage phosphor, absorption of x rays leads to the creation of trapped electrons at special sites in the bulk of the material. This trapped electron distribution is the latent image. Through optical

excitation, usually with a moving laser beam that stimulates the SP screen point by point, these trapped electrons can be freed from their traps, releasing their excess energy as light that can be detected with a photodetector and converted into an electrical signal. Devices that use this technology are available (eg, Agfa ADC Compact Plus and Solo, Kodak DirectView CR 850 and 950 scanners, Fuji FCR series of scanners). There are also storage phosphor scanners in development (Agfa, Fuji) that abandon the point-by-point method of reading and instead read an entire line of data points at once, enabling considerably faster CR systems than those commercially available today. These technologies will be described in more detail in the section on “New CR Developments.”

Actually, two modes of indirect imaging operation are possible with storage phosphors: up-conversion and down-conversion (Fig 4). In down-conversion, the high-energy (eg, x-ray) aerial image that exposes the SP screen creates a latent image proportional to the local exposure. Readout is accomplished by stimulating the screen with a constant (ie, flat-field) signal at some wavelength,  $\lambda_s$ , which causes the latent image to be emitted as a corresponding lower-energy

(visible) light signal at a different wavelength,  $\lambda_e$ . This is the mode used in today's medical CR imaging systems. In up-conversion, the high-energy radiation field that initially exposes the SP detector is constant (ie, it is not an image) and serves, in effect, to "charge up" or energize the detector uniformly. The image information arrives as a low-energy (eg, infrared) aerial image. This aerial image, with wavelength  $\lambda_s$ , stimulates the screen, which "discharges" the detector imagewise and causes it to emit a higher-energy visible-light signal at a different wavelength,  $\lambda_e$ .

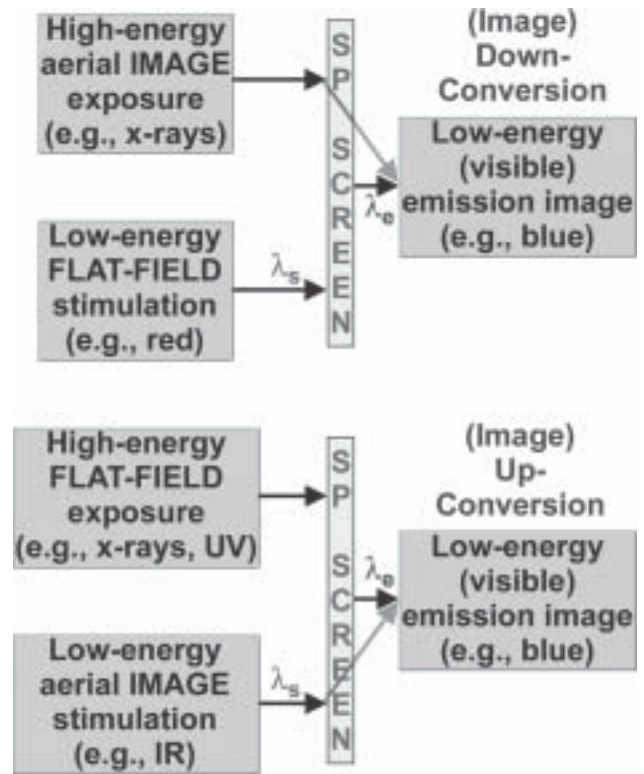
### A LITTLE HISTORY

Most people are familiar with the rapid development of CR during the past 20 years or so, during which the installed base of scanners has increased by a factor of about 20,000, system prices and sizes have each dropped by more than a factor of 10, and scanning speeds have increased by a factor of two to three. (Why speed has not kept pace with the other system characteristics will be addressed subsequently.)

However, storage phosphor technology has a much longer history than that (1). The PSL effect—that is, the storage of incident higher-energy radiation, with later release as visible luminescence through photostimulation—was used as early as the mid-1800s to convert invisible (eg, ultraviolet) aerial images into visible ones by means of full-field or area illumination. Not long after Roentgen's discovery of x rays in 1895, there were even experiments with full-field x-ray imaging that used PSL intermediates. As noted previously, these applications are examples of the down-conversion mode of operation, because high-energy (ultraviolet, x-ray) images are converted into visible images.

During World War II, infrared-stimulable storage phosphors were used in night-vision cameras, in which an infrared scene (the photostimulation source) imaged onto a previously energized SP detector would cause it to release its energy as a visible-light replica of the invisible input. This is an example of the up-conversion mode of operation, because the infrared image is converted by the storage phosphor into a visible image.

The real forerunners to the CR systems of today, however, were developed in the 1970s, when researchers started to look for ways to improve on the inefficient light collection and the resultant suboptimal image quality produced with the full-field illumination method. Their attempts led to the development of scanned storage phosphor systems (7), in which a focused beam of light stimulated the SP detector point by point, and a photodetector was placed close to the stimulation point to collect as much of the locally emitted luminescence as possible. These efforts culminated in the introduction of the first commercial CR system in 1981 (by Fuji Photo Film Co). Since that



**Figure 4.** Conversion modes available in storage phosphor imaging. In down-conversion, a high-energy (eg, x-ray) aerial image creates a latent-image signal proportional to the locally absorbed exposure. Subsequently, a constant (ie, flat-field) stimulation signal releases the stored latent image as a light signal proportional to the stored latent image. In up-conversion, the detector is first "charged up" or energized with a constant high-energy exposure, and the stimulation signal is a low-energy aerial image that "discharges" the constant latent-image signal imagewise. *IR* = infrared, *SP* = storage phosphor, *UV* = ultraviolet,  $\lambda_e$  = emitted wavelength,  $\lambda_s$  = stimulating wavelength.

time, numerous manufacturers have researched and produced commercial systems that use the PSL effect, and not just for medical imaging.

### CR VERSUS SCREEN-FILM SYSTEMS: THE BASICS

Although not shown in Figure 3, conventional analog screen-film (S/F) systems, which have been around for more than 100 years, also acquire images indirectly (1). A luminescent intensifying screen, operating in down-conversion mode, absorbs x rays and promptly emits light, which is detected by a piece of film in intimate contact with the screen. Chemical development of the latent image in the film produces more blackness (optical density) in areas that received more light and less blackness in areas with less light exposure. In S/F systems, the film is not only the acquisition medium. The combination of the film's characteristic curve (optical density vs light exposure) and the chemical development conditions provides the image processing. The film is also the display medium and the storage medium. This multifunctional role of film



**Figure 5.** Cross-section of a prototypic storage phosphor screen showing the two main layers (active phosphor and support). A number of other (manufacturer-dependent) layers are added to optimize the mechanical, optical, and electrical performance of the screen.

in the imaging system forces a number of design trade-offs, the most familiar of which are (a) x-ray exposure latitude versus display contrast and (b) sensitivity versus noise. This is the primary reason why there are so many different types of S/F systems on the market: the design trade-offs are different for different clinical applications.

In a CR system, the image acquisition stage also uses two components: (a) the screen, the detector that interacts with the x-ray aerial image to produce a (stored) latent image; and (b) the scanner, the readout device that extracts the latent image from the screen and converts it into a digital image. As we saw earlier, in a digital system, additional components are always required to perform the other necessary image-related functions. This separation of the five basic functions (acquire, process, reproduce, store, distribute) is one of the key advantages of digital systems over their analog competitors, because the individual stages can be optimized independently. Although the screen and scanner are discussed separately in the subsequent paragraphs, it is important to remember that they form a system. Changes in screen properties may drive different design decisions or trade-offs on the scanner side, and vice versa.

In many ways, storage phosphor screens are similar to the luminescent intensifying screens used in conventional S/F radiography. In both systems, the phosphor screens function as the primary x-ray absorber. Both types of screens also emit light promptly on exposure to radiation (x-ray-induced luminescence). With storage phosphor screens, however, this prompt emission is simply wasted, because only the stored latent-image signal is read out later. By contrast, in S/F radiography, this prompt emission is the only image signal available to the recording medium, namely, the film. Much research has gone into trying to reduce the prompt emission in SP screens, with the goal of retaining more stored signal for later readout. Unfortunately, the PSL in the storage phosphors of today occurs through the same pathways as the prompt emission, so that decreasing one will tend to decrease the other. In the CR systems of today, about half of the potential stored signal is still lost to prompt emission during x-ray exposure.

On a microscopic level, the physical structure of the screens in both technologies is also similar, consisting of small phosphor grains suspended in a binder material. In addition, the screens in both CR and S/F systems are reusable for many thousands of x-ray exposures; physical wear, not exposure-related change, is the primary indication for screen replacement in both systems.

There are also important differences between the technologies. Probably the biggest difference, and the most important advantage of CR over S/F systems, is exposure latitude. Film latitude is usually limited to a fairly narrow exposure range of about 30–40:1 by photon detection threshold and signal saturation effects (the so-called toe and shoulder of the characteristic response curve of the film). By comparison, the exposure limits of CR are about 10,000:1. The high tolerance of CR for exposure variation makes retakes because of incorrect exposure almost a thing of the past, which can decrease radiation dose to the patient. This exposure tolerance, however, is a two-edged sword. It can also mask systemic problems related to equipment malfunction or poor radiographic technique (eg, too high a dose) that would be immediately obvious with S/F systems. CR systems usually contain special software to help users detect and monitor such problems.

From the point of view of work flow and data flow, CR scanners can be placed in a central location, analogous to the centralized high-volume chemical processors in a typical screen-film environment. However, they also can be distributed geographically closer to where the images are acquired (ie, the examination rooms). This is an important work-flow advantage because image generation can continue while the digital image data are distributed over electronic networks to where they are needed. In a full-PACS scenario, in which images are interpreted in soft-copy mode and stored electronically, the CR scanners are simply modalities on the enterprise network.

## FORM AND FUNCTION IN CR: SP SCREENS

Conceptually, a modern storage phosphor screen consists of an active layer coated onto a rigid or flexible support (Fig 5). The active layer is the site of x-ray absorption, creation and storage of the latent image, and stimulated emission, while the support (eg, aluminum, glass, or polyethylene terephthalate [PET]) provides a smooth, sturdy surface for the sensitive phosphor layer, contributes to optical performance, and allows the screen to be handled and transported by both users and the CR scanner. The active layer, the thickness of which is usually adapted to the intended clinical application, contains small irregular phosphor particles (3–10  $\mu\text{m}$ ) suspended in a binder material.

In practice, a modern SP screen contains a number of additional (manufacturer-dependent) layers that

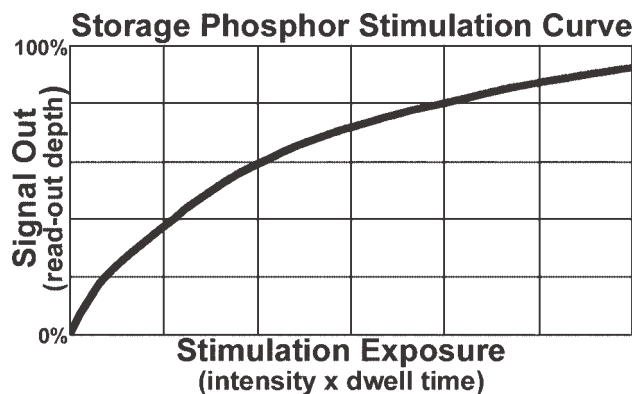
are needed to optimize its performance for clinical use (1,2,8). Mechanically, screens must be robust to user and machine handling, so special backing layers and overcoats are frequently used. Electrically, screens must be insensitive to electrostatic discharge (ESD), which can be addressed through a conductive layer on the screen surface.

Optically, screens must be optimized to enable as much as possible of the emitted light to escape to the screen surface, where it can be detected, while at the same time containing the spread of stimulating light (to maintain sharpness). These two apparently conflicting goals have produced a number of different solutions. In one, a black backing layer between the active layer and the support absorbs light, which reduces both the spread of stimulating light and the escape of the emitted light. In another solution, a reflective backing aids the escape of emitted light but also causes the stimulating light to spread more. An alternate solution, which is used in some newer screens, is to add a colored layer at the interface between the active layer and the support; this colored layer absorbs the stimulating light but still reflects the emitted light. Some manufacturers also disperse into the active layer a small amount of dye that preferentially absorbs the stimulating light to keep it from spreading too much.

Finally, to maintain high image quality, screens must also be insensitive to x rays that pass through or around the screen and are backscattered from objects behind it. This has led to the use of thin layers of lead (Pb), either in the cassette for the screen or on the screen itself (only with rigid supports). As a result, the simple two-layer screen concept becomes, in reality, a complex layer structure incorporating many design trade-offs and tuned to the characteristics of the scanning system.

Although many phosphor materials have been shown to have storage properties, the most commonly used screen materials in current commercial systems come from the barium fluorohalide family,  $\text{BaFX:Eu}^{2+}$ , where X is Cl, Br, I, or some combination thereof. In this chemical formula, europium (Eu) is called the activator. It is an impurity added in minute controlled amounts to the phosphor during the manufacturing process, and it strongly affects its storage properties and the spectrum of the emitted light. Other materials, such as  $\text{RbBr:Tl}^+$  and  $\text{CsBr:Eu}^{2+}$ , have been or are being used as storage phosphor materials in CR systems, and there is still much active research into finding new and optimizing current materials. Perhaps embarrassingly, despite the decades of research into SP materials and the PSL process, a number of mechanisms involved in their functioning at the microscopic level are still controversial or not completely understood. Work remains to be done!

On a more macroscopic level, the behavior of storage phosphors is much better understood (2). For ex-

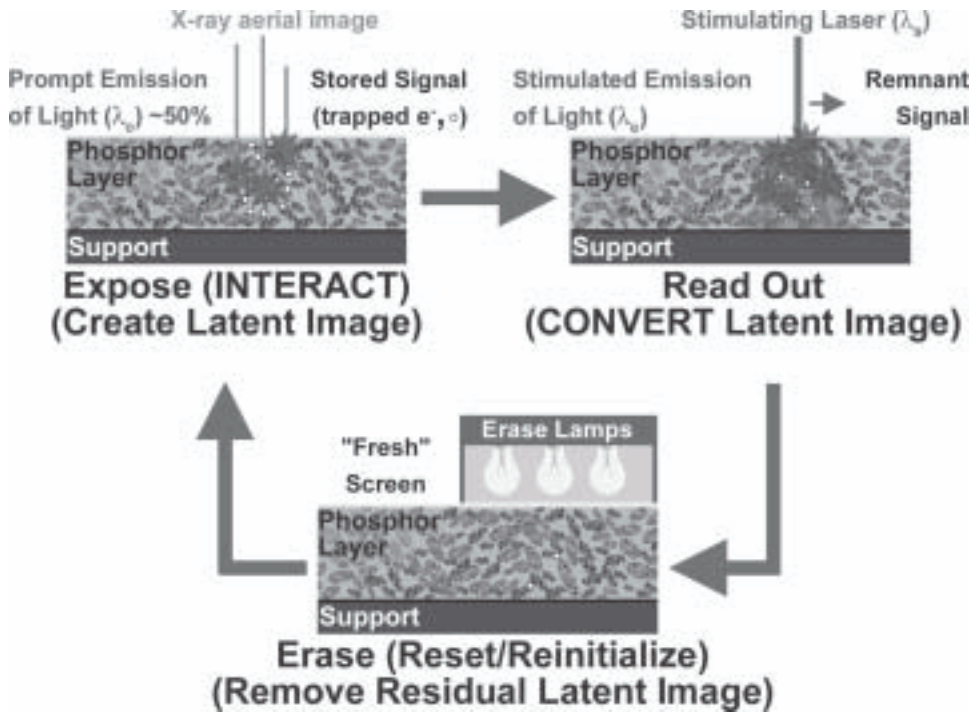


**Figure 6.** Stimulation curve of a typical storage phosphor. The amount of stored latent-image signal extracted from the screen increases nonlinearly with the amount of stimulating energy deposited. As the stored latent image becomes depleted, it becomes harder to extract the remaining signal, even with large amounts of deposited stimulation energy.

ample, when one looks at an SP screen, it appears white. This is due to the difference in the indices of refraction between the phosphor grains and the binder in the active layer, which creates an optically highly scattering or turbid medium. The consequences for the propagation of light in the layer are profound. Light entering the phosphor (eg, the stimulating laser light) or generated within the phosphor (eg, the photostimulated luminescence) spreads quickly in all directions from its point of entry or origin. In particular, a finely focused laser beam used to scan the latent image in the phosphor will broaden considerably as it moves deeper into the active layer. As a result, it will stimulate a larger volume of the stored latent image than that predicted by the diameter of the incident laser beam. The result is a loss of sharpness.

This phenomenon gives rise to the well-known trade-off between absorption and resolution. Thicker phosphor layers have better x-ray absorption (a desirable property) but also create greater unsharpness in the final image because of the increased spread of the stimulating beam (all other things being equal). One common solution to this trade-off is to produce at least two types of screens: a thinner screen for applications that require higher resolution (eg, mammography, imaging of extremities) and a thicker screen for applications in which x-ray absorption and dose are more critical than resolution (eg, thoracic and abdominal imaging).

Another interesting and important aspect of the PSL effect is the stimulation curve (Fig 6). Although one would like to stimulate as much of the stored latent image as possible out of the screen to obtain the best image quality, the amount of signal actually extracted depends on the total stimulation exposure to the screen, in other words, the total amount of energy deposited by the stimulation source. This, in turn, depends, to first order, on the intensity or power of the stimulating light and the



**Figure 7.** Three-step imaging cycle in CR. Cycle consists of the following: (a) an erased imaging plate, or screen, is exposed to x rays, creating a latent image within the screen; (b) the screen is stimulated (eg, with a moving laser beam) to emit light in proportion to the stored latent image; and (c) most of the remaining (unread) latent image is erased by using special high-intensity lamps, after which the plate is ready for the next exposure.  $e^-$  = electrons,  $\lambda_e$  = emitted wavelength,  $\lambda_s$  = stimulating wavelength.

length of time a given screen area is stimulated, which is also called the *dwell time*. However, this relationship is nonlinear, as the curve in Figure 6 shows. It is much easier to extract the first 50% of the latent-image signal than the last 50%. This screen property has a major influence on scanner design. The operating point of the scanner (eg, laser beam power and diameter, scan speed [dwell time]) will be a compromise among extracting as much signal as possible, scanning rapidly enough to satisfy user work-flow needs, and component costs, among other things. This dilemma will be addressed again in the "New CR Developments" section.

Like other DR technologies, the basic CR imaging cycle has three steps: (a) expose, (b) read out, and (c) erase, or reinitialize (Fig 7). Referring to Figure 7, in the exposure step, x rays are absorbed within the active layer, which creates a stored latent image of trapped electrons within the phosphor particles. For commonly used storage phosphors today, this latent image is relatively stable, lasting for many hours or even days before decaying to an unusable level. To read out the latent image, a focused laser is moved systematically across the surface of the screen, which frees the trapped electrons and allows them to return to a lower energy state, releasing their stored energy as a measurable light signal at a wavelength different from that of the laser. Note that, as already indicated, the laser beam expands as it moves through the screen, stimulating a volume that is considerably larger than that expected from the entrance diameter of the laser beam. A special erase unit containing high-intensity lamps removes any remaining (ie, unread) latent-image signal. This cycle can be repeated many thousands of times for each screen.

Note that the erasure step is not 100% effective; that is, the stored signal can never be erased completely. However, as long as the highest leftover stored signal on the screen after erasure remains well below the lowest expected signal from the next exposure, this residual "contamination" is unimportant. Even background radiation in the environment often causes an unwanted increase in the baseline signal of an erased screen, which appears as a noisy background in the image. This is why manufacturers recommend erasing screens shortly before their use, especially if they have not been used for a while.

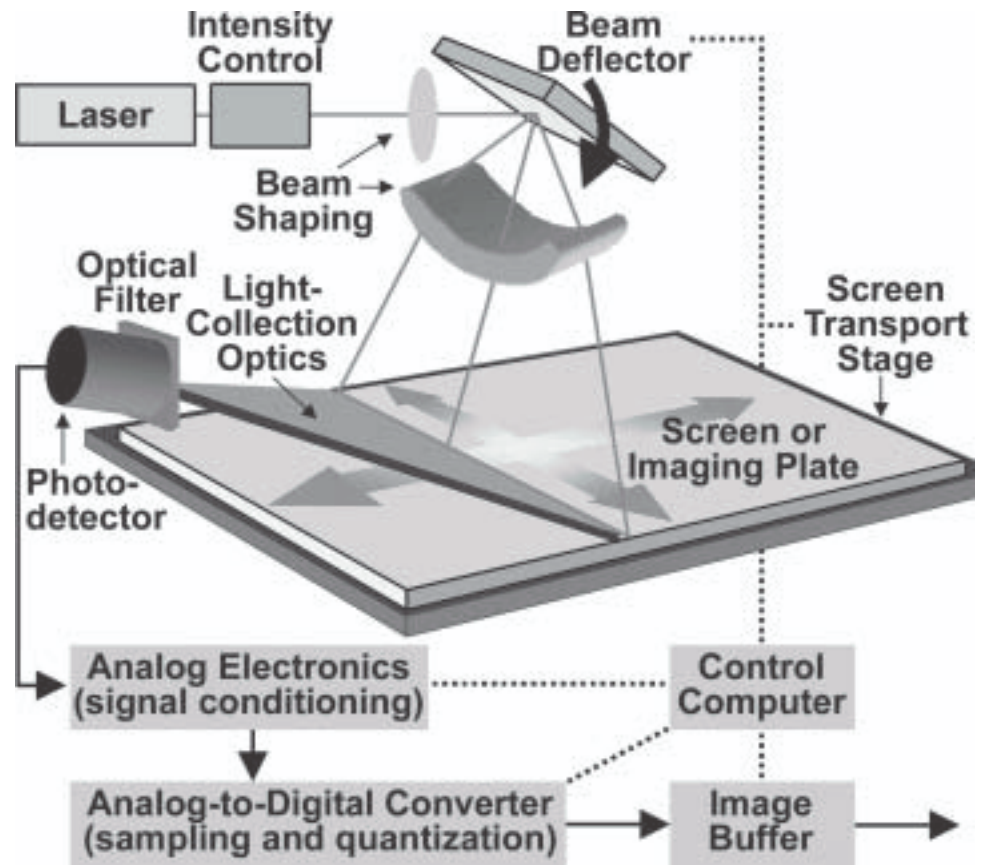
## FORM AND FUNCTION IN CR: SP SCANNERS

Current CR scanners all use a point-by-point readout technique in which a laser beam scans over the entire surface of the screen in some regular pattern (usually a raster scan), and the emitted light is measured at each point and converted into an electrical signal that is subsequently sampled and quantized into a digital image. In the earliest CR systems, the components to do this task almost filled an entire room. Today, they can fit onto a tabletop. With reference to Figure 8, this section will cover the main components of a typical CR scanner and some of the design issues involved (1,2,8).

### Laser Source and Intensity Control

Early CR systems used gas lasers (eg, HeNe, helium-neon laser with  $\lambda_s \approx 633$  nm in the red) to stimulate the trapped electrons in the screen. Most modern CR systems use red-emitting solid-state laser

**Figure 8.** Main components of a flying-spot CR scanner (see text for details).



diodes ( $\lambda_s \approx 670\text{--}690\text{ nm}$ ) as the light source. Red wavelengths are well matched to the stimulation spectrum of commonly used barium fluorohalide screens and are different enough from the emitted wavelengths (in the blue) that they do not interfere with detection. Solid-state laser sources are more compact, efficient, and reliable, and they last longer than gas lasers. Tens of milliwatts are easily achievable and are also necessary to extract as much signal as possible in clinically acceptable scan times. Laser beam dwell times for modern scanners are in the range from 1 to 6  $\mu\text{sec}$  per pixel.

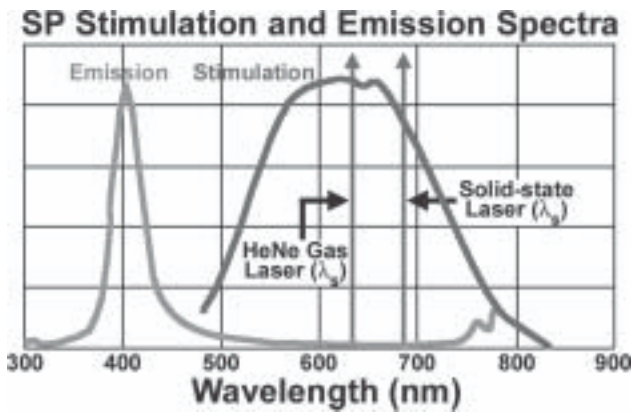
Not only do you need high laser powers for CR scanners, you also need constant power. Fluctuations in laser power translate directly (through the stimulation curve of the screen [Fig 6]) into fluctuations in output signal. In other words, a fluctuating laser intensity can make a constant latent-image signal (a flat field) appear to have structure. CR scanners incorporate special intensity controls that actively monitor laser power and correct such fluctuations. The tolerances are small. In the linear portion of the stimulation curve, fluctuations as low as a few tenths of a percent can create problems, so the intensity control must maintain the variability well below this level. Higher on the stimulation curve, the tolerance is not as high because a bigger change in exposure is required to produce the same change in output signal.

### Beam-shaping Optics

The beam that leaves the laser must be optimized for exposing the screen. This is especially true for solid-state laser diodes, which naturally produce an elliptical beam profile, rather than the circular ones produced by gas lasers. In addition, even with a circular beam leaving the laser, the beam will change shape and speed as it moves across the screen. You can simulate this effect by taking a flashlight with a roughly circular beam and moving it along a wall in front of you. Shine the beam perpendicular to the wall, and note that the beam is roughly circular. Then move the beam left and right along the wall, and note that its shape and its speed change with position. The beam gets more elliptical and moves faster as it travels farther along the wall away from the perpendicular.

In a CR scanner, this effect results in the stimulation of different screen volumes with different dwell times (scan speeds), depending on beam position. The effect is undesirable, because it means that even a uniformly exposed screen (ie, a constant latent image) would produce a different signal output and a different spatial resolution at the edges of the screen than in the middle. CR scanners include special optics (including one called an f-theta lens) that can keep the beam size, shape, and speed largely independent of the beam position.





**Figure 9.** Emission and stimulation spectra of a modern storage phosphor screen. Two laser stimulation lines, a helium-neon gas laser and a solid-state laser diode, are also shown. The separation of the emitted light wavelengths from the stimulating wavelength ( $\lambda_s$ ) is crucial to the function of CR.

### Beam Deflector

The beam deflector moves the laser beam rapidly back and forth across the screen to stimulate sequentially each point along a scan line. Motion in the other direction is handled by the transport stage (see next section). This direction is usually called the “fast scan” or “in-line” direction. Depending on the required scan speed, different kinds of deflectors can be used. For lower scan speeds, a rotating drum and a fixed beam (ie, no deflector; all motion handled by the drum) can be used. For somewhat higher speeds, the usual solution is a mirror mounted on a galvanometer. The galvanometer is made to oscillate back and forth, moving the beam across the screen. During its retrace, the laser light is blocked. For the highest speeds, a rotating polygon with multiple mirrors is used. Each mirror paints one line across the screen and then “hands off” the next line to the next mirror facet, and so on. Here, it is important that each mirror have the same reflectivity and same angle to the rotor of the polygon.

In all cases, the accuracy of the beam placement is critical; the scanner must be able to position the beam reproducibly at exactly the same point, line to line and scan to scan. The beam deflector must be able to position the beam consistently in the subpixel range to avoid artifacts like optical banding and “jaggies” (an artifact that makes straight edges look wavy or jittery).

### Transport Stage

The transport stage moves the image plate (IP) in a direction (roughly) perpendicular to the fast-scan direction. This direction is often called the “slow-scan,” the “page-scan,” or the “cross-line” direction. Between the actions of the beam deflector and the transport stage, the entire screen surface can be “touched” (ie, sampled) by the laser beam. As with the beam deflector, there are transport stage choices depending on the

speed requirements. A rotating drum can be used for slower scan speeds, but practically all CR scanners today use a linear transport in which the screen sits on a moving table or is clamped and moved along a track. In the case of cassetteless systems with integrated (nonremovable) imaging plates, the screens are attached to a conveyor belt that moves them to the appropriate processing station (exposure, readout, erase).

Here, too, the constancy of velocity is critical to avoid banding artifacts. Because the readout process is destructive, that is, the latent image disappears as it is read, there must be a consistent, constant overlap of the laser beam profiles across lines in the slow-scan direction. Fluctuations in transport velocity of even a few tenths of a percent can result in visible banding artifacts.

### Light Collection Optics

The light collection optics are used to collect as much as possible of the emitted light from the screen and channel it with minimal loss to the photodetector, where it is converted into an electrical signal. The image quality (signal-to-noise-ratio) is critically dependent on this step. Although the incident laser beam is highly directional, the turbid nature of the screen makes the emitted light come out in all directions. Thus, the light collection optics must sit close to the screen surface to intercept as many emitted light photons as possible. Manufacturers go to great lengths to devise clever designs that maximize the light collection efficiency. Some current CR systems use acrylic light pipes for this task. These pipes are wide and thin at the end close to the screen surface (covering the entire width of the screen), while the other end is tapered to fit the entrance aperture of the photodetector. Another design uses a highly reflective integrating cavity mounted over the screen that channels the light to photodetectors mounted along its length or on its ends. Fiberoptic solutions have also been proposed.

### Optical Filter

Without this seemingly innocuous component, CR would not work. Extracting a usable signal from a CR screen is only possible because the emitted light comes out at a wavelength different from that of the stimulating light. This spectral separation is critical (Fig 9). Even more critical, however, is allowing only the emitted light to enter the photodetector, which often has a fairly broad spectral sensitivity. It is relatively straightforward to calculate that the emitted light from a typical storage phosphor screen is roughly eight orders of magnitude (ie,  $10^8$ ) less intense than that of the stimulating light. Detecting the emitted light photons among the stimulating light photons is like looking for the proverbial needle in a haystack (in fact, if one assumes a 1-m-high hemispherical haystack and a standard sewing needle, the detection problem is comparable). The

optical filter plays the crucial role in keeping the stimulating light from entering the photodetector and swamping the desired image signal.

### Photodetector

The photodetector converts the emitted light photons into an electrical signal that can be processed into a digital image. Because of the low emitted light levels involved in CR, most commercial systems today use one or more photomultiplier tubes (PMTs). PMTs have high signal gain, reasonable quantum conversion efficiency (approximately 25%), low internal noise, and low dark current. In addition, their detection dynamic range is well matched to the range of signals possible from an SP screen in normal clinical use. A useful feature of PMTs is that their sensitivity in the red region of the spectrum is poor, so they effectively act as an additional filter for removing the stimulating light from the detected signal.

Because charge-coupled devices (CCDs) are less expensive and about twice as efficient as PMTs at converting light photons into electrical current, they are beginning to be used in CR systems. However, they also have some disadvantages that require more careful electronic and optical design. Optically, CCDs have a broader spectral sensitivity than PMTs, so they are also sensitive to the wavelengths of the stimulating light. This puts an additional burden on the design of the optical filter to remove the unwanted stimulation signal. Electronically, the dynamic range of CCDs is usually less than that of PMTs, and their internal noise and dark current levels are somewhat higher, so careful circuit design and high-quality electronics are needed to compensate. Despite these additional design constraints, the lower cost, smaller size, and flexibility of CCDs will draw them more and more into the CR mainstream for optical detection.

### Analog Electronics

The signal emerging from the photodetector is an analog signal that represents the variations present in the screen's latent image and, therefore, in the original x-ray exposure. Unfortunately, the exposure range of any given medical image can be high and is often not known a priori. Early CR systems addressed this problem by performing an optical prescan. This was a scan of the screen at low laser power with a defocused laser spot to see what range of signals could be expected during the subsequent full-power scan. Modern systems address the problem electronically.

One way to ease the design requirements on the electronic chain is to compress the analog image data before digitization. This means that the input signal (ie, the signal leaving the photodetector) is mapped nonlinearly into a new quantity that varies less than the input signal. (This also has some advantages for

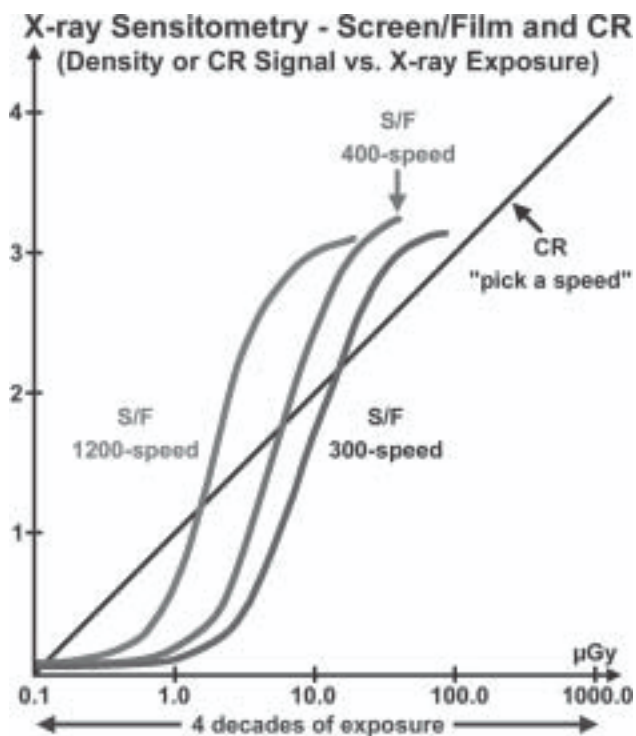
later human viewing that are beyond the scope of this chapter.) Most commonly, analog, logarithmic compression is done. The photodetector signal is boosted with a logarithmic amplifier before being sent to the analog-to-digital converter (ADC). An alternate compression technique is based on a square-root amplifier, which has the advantage of equalizing the quantum noise inherent in the exposure while it compresses. Other manufacturers elect to process the input signal linearly (ie, uncompressed) and then compress the digitized data later by selecting a subset of digital gray levels for the final image. All of these methods produce usable results, but if a user wants to convert digitized values back into original x-ray exposure values (eg, for image analysis), then knowledge of the compression scheme is important.

Another aspect handled by the analog electronics following the photodetector is preparation for the sampling process. The basic rule of sampling (Nyquist theorem) states that for proper undistorted digitization of an analog input signal, the sampling frequency (ie, how often the signal is measured along the scan line) must be at least twice the highest frequency present in the input signal. The signals present in the photodetector signal contain a broad range of frequencies (including noise), but some are not needed for diagnosis or compatible with the digitizer. Therefore, manufacturers include so-called antialiasing filters in the analog chain to remove these higher frequencies before the ADC (aliasing is the distortion produced when the Nyquist theorem is not satisfied).

### Analog-to-Digital Converter (ADC)

The ADC is the boundary between the analog and digital worlds and encompasses two processes: sampling and quantization. The ADC works with the control electronics to produce a digital image that is at least equivalent, for purposes of the imaging application, to the analog image from which it stems. The movement of the laser beam across the screen converts spatial variations on the screen surface into a time-varying signal from the photodetector. This time-varying signal must be sampled at a high enough rate to preserve the level of spatial resolution needed for the application.

Similarly, the intensity variations of the photodetector signal must also be sampled or quantized finely enough to preserve the signal variations (contrast) required for the application while still covering the entire potential exposure dynamic range. For example, data that have been compressed by using a logarithmic or square-root compressor typically receive 8–12 bits of quantization per pixel. For uncompressed (ie, linear) data, 12–16 bits per pixel are common. The criteria for successful digitization are dependent on the application; for example, the digitization needs for mammography are different from those for abdominal imaging.



**Figure 10.** Input/output relationship of a CR acquisition system. The CR system is linear over roughly four orders of magnitude (four decades) in exposure. Compare this with several screen-film (S/F) systems that cover the range of typical clinical exposure conditions but provide much more limited exposure latitude and must therefore be matched carefully to the exposure technique.

### Image Buffer

The digital images generated by the scanner need to be stored somewhere before they can be distributed to their final destinations (workstations, archives), so some local storage is generally available in the form of a hard disk drive. The capacity of this drive is matched to the throughput of the scanner and typically includes the capability to keep the scanner operating even if its connection to the network has been interrupted.

### Erase Station

Not shown in Figure 8 is the erase station used to remove any remnant signal from the screen and reinitialize the screen for the next exposure. This component typically consists of an array of high-intensity lamps, with intensities often several orders of magnitude higher than the stimulating light source, that drive the remnant signal down to a level considerably lower than the lowest expected exposure in the next image (it is never possible to remove all of the remnant signal). The duration of the erase cycle depends on the desired level of erasure, the intensity and emission spectrum of the lamps, and the erasability of the storage phosphor material. Note that natural background radiation can produce a noise field (a stored latent image) in the screen, so it is always advisable to erase before use any screens that have not been erased for a while.

## IMAGING PERFORMANCE OF CR

Many imaging performance metrics can be measured in CR and DR, including signal response (the input/output [I/O] relationship), resolution (typically expressed as a modulation transfer function [MTF]), noise (typically expressed as a noise power, or Wiener spectrum), and more advanced metrics, such as dose efficiency (exemplified by the detective quantum efficiency or DQE) (9,10). Because the detailed imaging performance of various DR technologies, including CR, is covered in another chapter of this syllabus (“Assessment of Display Quality” by Ehsan Samei, PhD), this chapter will focus only on some general performance issues related to the input/output relationship in CR (contrast, dynamic range), the factors affecting spatial resolution, and the factors affecting noise (1,2,8).

### Input/Output Relationship

CR is often described as an acquisition technology that is linear over four or five orders of magnitude, which is not too far from the truth. Storage phosphor screens, indeed, are linear detectors over more than five orders of magnitude. However, this impressive exposure dynamic range is throttled somewhat by the scanner. For example, high x-ray exposures can produce light signals that cause the photodetector to become nonlinear (eg, saturate; even PMTs have their limits). At the low-signal end, the reflection or scattering of stimulating light onto other low-signal portions of the screen during scanning, which is called flare, can make it impossible to detect latent-image signals below a certain level. In addition, various sources of noise (see “Noise” section) produce a “noise floor” below which no useful signal is detectable. Even with these limitations, CR systems provide a detection range that is more than capable of handling the exposures found in diagnostic radiography and is orders of magnitude broader than the screen-film systems they are intended to replace (Fig 10).

Despite more than 20 years of use and hundreds of publications, many people still confuse the concepts of latitude and dose. Such statements as “CR is a lower-dose technology than screen-film” or “we saved a factor of  $X$  in dose by switching from screen-film to CR” are patently false and indicate a fundamental misunderstanding of CR. The fact that you may be able to reduce dose with CR relative to your current screen-film system does not make CR a lower-dose system. One could have used a different, higher-speed S/F system and achieved exactly the same result. The fact that dose is reduced with CR relative to a specific S/F system has to do with its latitude, not its dose efficiency.

In fact, from a dose-efficiency point of view, CR and S/F systems are actually fairly comparable. Many studies have shown that to achieve the same objectively measured image quality as S/F systems, CR

Exposure-related Sources	Screen-related Sources	Scanner-related Sources
X-ray quantum noise	Fluctuations in number of x-ray quanta absorbed in screen	Deflector and transport stage velocity fluctuations
Equipment variability (eg, x-ray tube fluctuations)	Fluctuations in number of x-ray quanta scattered in screen	Beam intensity variations (from laser, intensity control)
	Number of trapped electrons created per absorbed x-ray quantum	Spread or scatter of stimulating beam
<b>Screen structure noise</b>	Variable latent image decay	Photostimulable luminescence photons collected by light collector
	Photostimulable luminescence photons emitted in screen	Electrons created in photodetector per photostimulable luminescence photons
	Photostimulable luminescence photons escaping screen	Analog electronics noise
	Active layer internal structure (phosphor particles, binder)	Analog-to-digital converter variability in sampling and quantization
	Overcoat or backing layer structure	Nonuniformities in erase process
	Phosphor particle size distribution	

Figure 11. Potential sources of noise in CR systems.

usually needs a bit more x-ray exposure (remember that about half of the potential signal in CR is already lost as prompt emission during the exposure). Given this parity, the fact that people are willing to operate a new CR system at a lower dose level than their current S/F system may indicate that (a) they are capable of lowering their diagnostic image quality standards (ie, the quality level needed for the diagnosis is not what they thought it was); (b) they have been unnecessarily overdosing their patients with their current S/F system (ie, a faster S/F system would have worked just as well); and/or (c) image processing (the subject of another chapter in this syllabus, "Processing Digital Radiographs of Specific Body Parts" by Flynn) has turned an objectively worse CR image into one that is subjectively more pleasing than their S/F image (and still contains the necessary diagnostic information). What this willingness does not indicate, however, is that CR has lower dose requirements than S/F systems. If dose were the overriding issue, the world would already be using 1,200- or 1,600-speed S/F systems, which is clearly not the case.

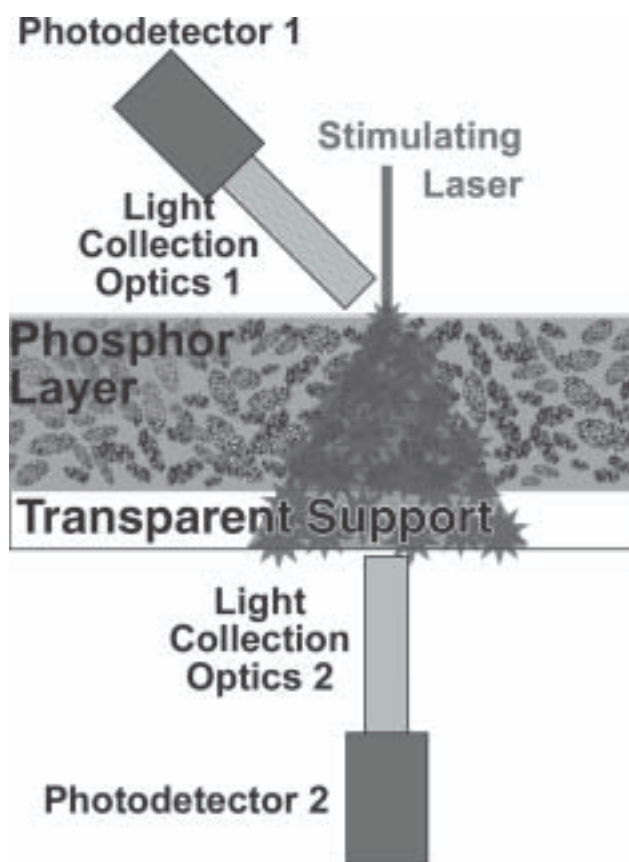
The crucial point is that with CR, one need not find a new detector to experiment with dose reduction. The same screen and scanner will work under a large variety of exposure conditions (a range of 10<sup>4</sup>) without changing anything. This is not true for screen-film systems, which are limited to about 30–40:1.

### Spatial Resolution

Many factors affect spatial resolution in CR. The most critical one, phosphor layer thickness, has already been described. In S/F systems, x-ray-induced luminescence created within the turbid phosphor layer spreads or scatters in all directions; and some of

the luminescence, after traversing the layer thickness, eventually exposes the film. In CR, the spread of the incident stimulating light within the turbid phosphor layer, the extent of which is largely determined by the layer thickness, is the prime determinant of spatial resolution. The trade-off between absorption and resolution was the compromise between detecting more x-ray quanta and maintaining sharpness, the result of which was a family of screens adapted to different applications.

Other factors also affect sharpness, however. Recall that CR system prices and sizes have decreased by a factor of about 10 over the last 20 years, while system speed has only doubled or tripled. The reason for this unspectacular speed increase is called afterglow. Storage phosphors continue to emit light for a short time after stimulation is stopped. The time constant for this material-dependent decay is called the luminescence decay time. For the materials commonly used in CR today, this time is around 0.7–0.8 μsec. As the laser beam moves across the screen, the light being collected from the current position can be mixed with the still "glowing" emission from the previous laser positions. This causes a one-dimensional smear or resolution loss in the output signal. At low scan speeds, this is not an issue, but as the dwell time per pixel (in the range of 1–6 μsec per pixel today) starts to approach the luminescence decay time, spatial resolution will be affected. Although there are image preprocessing techniques that can mathematically unscramble this effect, they do so only with a loss in signal-to-noise ratio (ie, image quality). Thus, the long-standing drive toward faster scanners has been throttled by a natural physical limit of existing phosphors.



**Figure 12.** Dual-sided reading that uses a storage phosphor with a transparent support. Signal is extracted from both sides of the screen by using dual light-collection–photodetector systems and the associated electronics.

A number of other notable factors also influence spatial resolution. The antialiasing filter mentioned earlier reduces the high-frequency content of the image to match the capabilities of the digitization system. The response of the filter is not perfect; that is, some of the diagnostically relevant spatial frequencies below the cutoff are also affected. With modern filter design tools, the effect on real image information can be minimized. The fact that SP readout is destructive also has an effect on spatial resolution (positive and negative), but the details are too far afield for this chapter (11).

### Noise

Noise is the random variation of some output signal around the mean value predicted by its I/O relationship. A list of potential noise sources for CR (Fig 11) can be subdivided into screen-related, scanner-related, and exposure-related sources. The exposure-related noise is primarily the quantum noise inherent in the x-ray beam. The screen-related sources can be broken down into factors related to the physical structure of the screen (eg, grain size distribution, supplemental layer structure) and performance-related factors (eg, x-ray absorption, scatter, conversion efficiency). Finally, practically every scanner component discussed previ-

ously is also a potential noise contributor—some more serious, some less. The list of noise sources for CR is somewhat longer than the comparable list for S/F systems, which is at least qualitatively consistent with the need for increased exposure in CR to reach a comparable level of objective image quality.

### NEW CR DEVELOPMENTS

CR has come a long way in about 20 years, from laboratory curiosity to mainstream imaging modality. Still, the proved and reliable flying-spot scanner with powder image plates is getting close to its fundamental physical limits with regard to factors such as image quality and scan speed. Improvements to the status quo must involve new approaches to CR, and a number of these exist. Some have either become commercially available recently or will soon enter the market. There is considerable potential for improvement in image quality and speed, to the point that future CR systems can remain competitive with the new flat-panel DR detectors entering the market. The main new developments are covered in the following sections.

#### Dual-sided Reading

The idea of detecting emitted light from both sides of the SP screen to extract more signal (and improve signal-to-noise ratio) has been around for a number of years (12), but commercial products that use this readout technique are relatively new to the market. In this development, the substrate of the screen is made transparent, and a second set of light collection optics, along with a photodetector and electronics, is added on the opposite side of the screen (Fig 12).

This configuration has several benefits. First, more emitted light signal can be collected without changing the dwell time per pixel. Second, combining the two detected signals in a spatial-frequency–dependent way creates a total output signal that has better signal and noise properties than either alone. Note, however, that the stimulating beam has broadened considerably by the time it reaches the back or bottom of the active layer, so the light signal exiting the bottom is considerably less sharp than that collected at the top. As a result, the image quality benefits more from combining the two signals at lower spatial frequencies (where both signals are made to contribute) than at higher ones (where the contribution of the back-screen signal is selectively reduced). An additional bonus is that one can increase the layer thickness a little to improve x-ray absorption without a marked loss of sharpness; this can be controlled by the signal combination parameters.

#### Structured Storage Phosphors (Needle Image Plates)

Structured phosphors (Fig 13), that is, phosphors with a nonisotropic physical structure, have been

around for a long time (2,5). They are used, for example, in image intensifier tubes to absorb x rays and channel the resultant x-ray-induced luminescence to subsequent stages in the imaging chain. Such “directional” phosphors also find use in the new indirect flat-panel DR systems (covered in a separate chapter of this syllabus, “Digital Radiographic Technology” by Yorkston). Rather than being coated onto a substrate like conventional phosphors and current powder-based storage phosphors, structured phosphors are grown under carefully controlled temperature, pressure, and mechanical conditions to form long crystalline rods or needles roughly perpendicular to the substrate.

The needle structure has a number of advantages. Most obviously, the needles tend to keep any luminescence generated inside them traveling along the needles, which helps to maintain image sharpness. In addition, because the needles are grown on the substrate, there is no need for a binder. This means that the active layer volume is almost entirely phosphor, leading to a major (about twofold) increase in x-ray absorption for the same active layer thickness. The active layer can also be made thicker (twofold is not uncommon) with little loss in image sharpness, which again leads to a major increase in x-ray absorption. This easing of the trade-off between absorption and resolution brings with it new flexibility in system design. Finally, the active layer in a structured phosphor is more spatially uniform than that in a powder-based phosphor, which can decrease screen structure noise.

Although the use of structured phosphors in image intensifier applications is decades old, it has been more difficult to find appropriate materials that demonstrate the PSL effect. One material,  $\text{RbBr:Tl}^+$  (13), was used in a dedicated chest system some years ago, but the material had a number of characteristics (lower x-ray absorption than  $\text{BaFX:Eu}^{2+}$ , rapid decay of the latent image, hygroscopicity) that limited its clinical usefulness. Recently, a promising material,  $\text{CsBr:Eu}^{2+}$ , has been found (14) that has good storage properties and can be grown in needles. Investigations showed that this material has a number of other attractive properties (higher x-ray absorption, conversion efficiency, stimulability, and erasability than  $\text{BaFX:Eu}^{2+}$ ) that could make it well suited for clinical use in future storage phosphor systems.

Initial evaluations of needle IPs based on  $\text{CsBr:Eu}^{2+}$  showed that the achievable objective image quality is considerably higher than that of conventional storage phosphor screens. In fact, the measured image quality can compete favorably with that of modern indirect flat-panel DR systems that use a CsI scintillator and an amorphous silicon TFT detector array. Observer performance comparisons between the needle IPs and conventional IPs have also shown the superiority of the structured screens in the detection of small low-



**Figure 13.** Cross-section of a structured SP screen showing the needlelike storage phosphor rods perpendicular to the substrate. The substrate can be either opaque or transparent, depending on the application and the desired scanning geometry. As with powder SP screens (Fig 5), additional layers are needed to optimize performance.

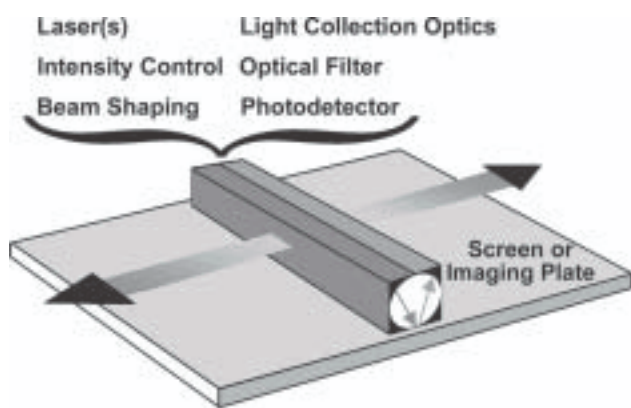
contrast objects. CsBr screens for CR are not yet available commercially.

### Line Scanning

Although systems have become about 10 times smaller over the last 20 years, modern flying-spot scanners still consist mostly of discrete components, which leads to a fairly low packing density. In addition, flying-spot scanners are subject to the “speed limit” of the IP’s luminescence decay time, which sets a minimum pixel dwell time to avoid blurring. A novel approach to read-out (15), namely, line scanning, addresses both of these deficits (Fig 14). In this approach, an entire line is illuminated with a set of stimulation sources, for example, a row of solid-state laser diodes, and the output signal from the whole line (or multiple lines) is read out by an array of photodetectors (eg, CCDs). The stimulation sources, beam-shaping optics, light collection optics, filters, and photodetectors are all contained in a scanning head that is as wide as the screen, and the entire screen surface is scanned through the relative linear movement of the head and the screen (eg, moving the scanning head over stationary image plate or vice versa).

The integration of multiple discrete components into a compact scanning head produces a much smaller CR system. More important, because there is no fast-scan motion, the luminescence decay time is no longer an issue. In a line scanner, the dwell time per pixel can be milliseconds rather than microseconds and still produce a faster scanner. For example, in one prototypic scanner with a pixel dwell time of 2 msec (three orders of magnitude longer than today’s flying-spot systems), scanning an IP into 2,500 lines takes only 5 seconds (faster than the fastest flying-spot CR scanner today).

An additional advantage is the proximity of the light-collection-photodetector subsystem to the emitted light, which improves overall collection efficiency. Finally, the photodetectors used (eg, CCDs) are more efficient than the PMTs used in most flying-spot scanners, which leads to an increase in detected signal. Manufacturing such line-scanning systems is difficult, however, and involves massive component miniatur-



**Figure 14.** Schematic diagram of a storage phosphor scanner based on the line-scanning principle. In this embodiment, the scanning head moves over a stationary screen. Many of the discrete components found in flying-spot scanners (Fig 8) are integrated into the scanning head. Depending on the design, even the analog electronics and analog-to-digital converter can be incorporated into the head.

ization, tight mechanical tolerances, and complex optical and electronic design. Prototypic line scanners have been shown to produce image quality results comparable to or slightly better than today's flying-spot scanners when used with powder IPs. When these prototypic line scanners are combined with needle IPs, the results have been comparable to those produced with indirect flat-panel DR systems.

### Other New Developments

There are also some interesting new developments on the applications front. One of these is dual-energy imaging (2). Dual-energy imaging has been around for many years, but the exposure and plate-handling procedures (ie, two separate exposures at different energies vs a single exposure of two detectors with an interposed metal filter) were not optimal. CR systems now on the market make the image acquisition step much easier. These single-shot CR systems have automated plate handling, built-in metal filters, and image registration software that allow reasonable throughput (about 40 exposures per hour) and work flow with much less effort.

The topic of CR for mammography has always been controversial. While CR was used routinely for every other clinical application, all but a few users (16) hesitated to use it for diagnostic or screening mammography because of fears that its limited spatial resolution relative to screen-film systems would lead to lower clinical performance. This situation is changing. New higher-resolution screens made especially for mammography, along with new high-resolution (but still flying-spot) scanner designs, including dual-sided reading (17), have enabled major progress on the mammography front. Combined with dedicated image-processing algorithms, mammography with CR has entered a new era, which has

been confirmed by the regulatory clearances that have been issued for new mammography systems with CR.

### SUMMARY

CR has developed over the last 20 or so years from a curious new technology that could not quite compete with screen-film systems to a commercially successful diagnostic workhorse that often displaces screen-film systems. Interestingly, the same statements made 10–15 years ago about the advent of CR leading to the disappearance of film are being reworded by some today to announce the disappearance of CR in the wake of new flat-panel DR systems. All three types of acquisition technologies currently coexist, however, and will continue to do so for many years. Moreover, new developments will bring CR image quality to a level competitive with that of these new DR systems at a considerably lower cost. For some applications, the throughput and work flow of flat-panel DR systems will offer clear advantages over CR, while for others, particularly those applications that require cassette-based operation (eg, portable radiography), CR will still have advantages. The ability to network multiple in-room or centralized CR scanners also provides flexibility in work flow and throughput, not to mention redundancy in the event of system downtime. So, although CR might be labeled by some as old and familiar technology, it is actually still dynamic and has the potential for considerable improvement and optimization in the future.

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