Digital radiography: CR versus DR? Time to reconsider the options, the definitions, and current capabilities

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Categorizing digital radiography systems is no longer as simple as "computed radiography" (CR) and "direct and/or digital radiography" (DR). The more technology changes, the more it changes. The CR versus DR issue is no exception, where historical boundaries of CR (with passive, cassette-based image acquisition and detector handling with offline processing) and the fully integrated DR system (with automatic processing and display) are now converging. Historically, CR has referred to the implementation of a cassette-based photostimulable storage phosphor (PSP) imaging plate reader and quality-control workstation that are packaged as an add-on system to existing X-ray devices using screen-film detectors. In contrast, DR has been touted as a totally integrated X-ray source-generator-detector solution with images displayed for review within the room shortly after the X-ray exposure. Now, "CR" technology (specifically using PSP converters) has been developed into integrated X-ray systems and has completely automated acquisition, display, and processing. Now "DR" technology has cassette-based detectors available for situations requiring more flexibility in positioning and multiple uses in conventional X-ray cassette trays and bedside portable radiography.

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Computed radiography uses a PSP that transiently stores a latent image in the form of electrons in semistable traps within the phosphor structure, with subsequent data extraction using a stimulating point laser beam that scans the phosphor surface point by point. Line scan laser sources coupled to an array of microlens and photodiode light detectors stimulate and acquire data in parallel, reducing the readout time of a PSP imaging plate from about a minute to ≤10 seconds, which is comparable to other digital radiography detectors known for readout speed.

In its current definition, *DR* describes a multitude of digital X-ray detection systems that immediately process the absorbed X-ray signal after exposure and produce the image for viewing with no further user interaction. This fact has resulted in the use of the term *direct radiography* by many manufacturers, which has added to the nomenclature confusion, because of another use of *direct* that describes the conversion of X-rays into charge, which is described later. In terms of CR versus DR, inclusion of automated CR systems would fall into the DR category, along with optically coupled scintillator to charge-coupled device (CCD) camera systems, fiberoptically coupled rectangular CCD array slot-scan detector systems, complementary metal-oxide semiconductor (CMOS) detector systems, thin-film-transistor (TFT) flat-panel detector systems, and slot-scan photon counting detectors, which were recently introduced. The *CR* and *DR* acronyms no longer define the essence of digital radiography detector attributes, since distinct classification into these 2 broad categories is no longer possible and, in fact, often leads to confusion and "marketeering" claims and counterclaims.

In a recent publication by a multisocietal effort (including the American College of Radiology, the American Association of Physicists in Medicine, and the Society of Imaging Informatics in Medicine) to determine practice guidelines for the use of CR and DR in the clinical environment,[1] a consensus to title the work *Practice Guidelines for Digital Radiography*[2] was made to be inclusive of all types of digital radiographic systems, in recognition of the issues explained above. Likewise, in this article, the term *digital radiography* refers to all types of digital radiographic systems, including both those that had been historically termed *CR* and those historically termed *DR*.

So, how should one categorize current state-of-the-art digital radiography technology? One method promoted in this article considers 3 characteristics: 1) detector form factor; 2) image acquisition

Table 1. Digital radiography technology attributes. Categories include X-ray to signal conversion methods, conversion materials, signal coupling methods, and detector form factor, including associated devices.

Technology	X-ray to signal	Conversion materials	Coupling methods	Detector form factor
PSP	Indirect: electron trapping and photostimulated luminescence	• BaFBr (unstructured)	Light guide with PMT	Cassette + PSP reader Integrated mechanical plate changer
		• CsBr (structured)	Microlens array and CCD	Cassette + PSP reader Integrated cassetteless
Flat-panel *a*-Si TFT array	Indirect, scintillator Integrated cassetteless	• Gd_2O_2S (unstructured)	Photodiode + TFT array	Wired or wireless cassette
		• CsI (structured)	Photodiode + TFT array	Integrated cassetteless
	Direct, semi-conductor	• *a*-Se	HV electrodes + TFT array	Integrated cassetteless
CCD	Indirect, scintillator	• Gd_2O_2S (unstructured)	Optical lens	Integrated cassetteless
		• CsI (structured)	Optical lens	Integrated cassetteless
		• CsI (structured)	Fiberoptical	Mechanical slot-scan
CMOS crystalline Si	Indirect, scintillator	CsI (structured)	Photodiode + CMOS array	Wired or wireless cassette Integrated cassetteless
Photon counters	Direct	• Xe gas	HV electrode + antenna array	Mechanical slot-scan
		• Solid state Si	Amplifier circuits	Mechanical slot-scan

TFT= thin-film transistor; PSP = photostimulable storage phosphor; *a*-Si = amorphous silicon; CMOS = complementary metal-oxide semiconductor; CCD = charge-coupled device; BaFBr = barium fluorobromide; Si = silicon; CsBr = cesium bromide; CsI = cesium iodide; *a*-Se = amorphous selenium; Gd_2O_2S = gadolinium oxysulfide; Xe = xenon.

geometry; and 3) X-ray signal conversion method, as explained below. Various digital radiography detectors and attributes are listed in Table 1.

Detector form factor considers aspects such as "cassette" versus "cassetteless" (Figure 1) and "passive" versus "active" operation. A majority of cassette-based digital detectors are based on PSP technology as a direct replacement for screen-film, thus providing a cost-effective means to achieve digital image acquisition, while at the same time allowing great positioning flexibility. Passive operation allows an asynchronous coupling of the X-ray exposure and storage of the image signal on the phosphor. Subsequent processing is performed by physically inserting the cassette/phosphor plate into an image "reader" to render the radiograph. Multiplexing of several rooms to 1 reader is possible but is also time-inefficient and potentially limiting in busy, high-throughput rooms. An alternative is a cassette-based TFT-array detector for use with portable radiography examinations and as a "drop-in" to cassette trays in general radiographic rooms. This detector uses a wired electronic connection to the X-ray generator to actively integrate and read out the X-ray image after exposure. A CMOS detector in a cassette configuration for mammography is available as a replacement for screen-film mammography. Wireless interfaces will likely soon appear on such detector systems. Image display shortly after the exposure and good positioning flexibility are benefits; damage to the detector by unintentional mishandling is a potentially costly drawback.

Cassetteless digital detectors are part of an integrated X-ray generator, X-ray tube, and detector system. The basic meaning of "cassetteless" indicates that the image is acquired and subsequently displayed at the in-room technologist console for review and manipulation with minimal user interaction. The earliest "cassetteless" systems had PSP plate changers with automated acquisition, point-scan laser readout, image processing, and display, with individual processing times on the order of a minute. Recent introduction of line-scan laser excitation of the PSP imaging plate reduces readout time of a large 35 × 43-cm field-of-view (FOV) detector in as few as 5 seconds, with automatic processing and display. This is comparable to many large FOV optically coupled CCD and flat-panel TFT arrays, which comprise the majority of cassetteless digital radiography systems. In the extreme are "real-time" flat-panel systems specifically manufactured for fluoroscopic imaging that also provide radiographic capabilities. Other cassetteless radiographic systems include slot-scan acquisition devices in which the image is produced by scanning a collimated beam and detector array across the FOV over a time span of seconds.

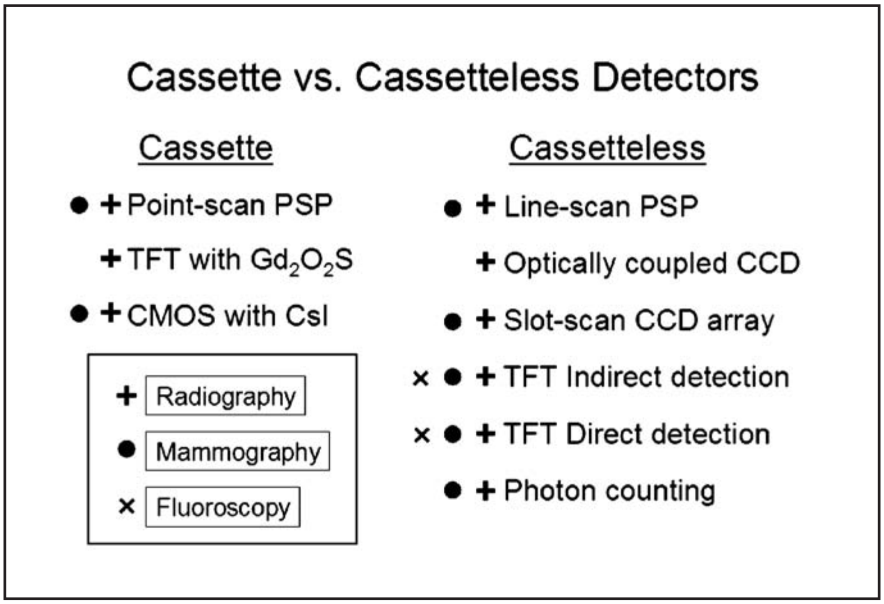

FIGURE 1. *Digital radiography detectors can be categorized into "cassette" and "cassetteless" in terms of form-factor and functional operation. The inset legend indicates the applications for each of these DR system technologies. "CR" (PSP) and "DR" are represented in both areas. TFT = thin-film transistor; PSP = photostimulable storage phosphor; CMOS = complementary metal-oxide semiconductor; CCD = charge-coupled device; CsI = cesium iodide;* Gd_2O_2S *= gadolinium oxysulfide.*

Image acquisition geometry distinguishes instantaneous acquisition of a large-area digital image with a short exposure time versus a sequential, long exposure time acquisition of a slot-scan device. Most digital detectors for radiography use a large-area FOV geometry, which allows short exposure times to decrease the probability of patient motion. However, detected scatter from the patient reduces image contrast and consumes a fraction of the digital range of the detector with essentially useless information. Thus, an antiscatter grid is often used with thicker body parts (it is used in essentially all adult imaging except extremity examinations) and incurs a relatively large dose penalty (typically double the patient dose) because of the loss of primary radiation. The slot-scan geometry with pre- and postpatient collimation produces a narrow fan-beam incident on a scintillator coupled to a time-delay-integrate CCD photodetector array.[3] Since at any instant only a small fraction of the patient volume is irradiated, the amount of scatter is reduced to low levels, and the amount of detected scatter is extremely low because of postpatient collimation and a small detector area. Grids are not required, which results in good dose utilization; image acquisition times, however, are extended into seconds, and the detector/collimator configuration makes positioning a challenge for nonstandard imaging protocols.

X-ray signal conversion is described by indirect, direct, or photon-counting methods. All digital detectors produce an output signal, usually in the form of electrons or holes (positive ions), which represent a quantity of charge that is proportional to the number of X-rays absorbed at a specific detector element (del) position. The magnitude of the charge is converted to a voltage and then to a digital value for storage in the image matrix at the corresponding del location in the detector plane.

Indirect refers to the conversion of X-rays into secondary information carriers, such as absorbed X-ray energy to light energy conversion in a scintillator, or to stored electrons in semistable traps within a storage phosphor and subsequent stimulation with a laser and light emission. For each X-ray photon absorbed, a large amplification of light photon carriers is produced on account of the large energy difference of X-rays (keV) versus light (eV) or energy required to trap electrons in a storage phosphor (eV). Regardless of the method by which secondary light photons are formed, the large numbers are directed to a wavelength-matched sensitive photodiode-TFT array via direct optical coupling, to a CCD area detector by lens-optical coupling, or via a light-guide pickup to a photomultiplier tube in a PSP reader device. Charge amplitude is generated in response to the amplitude of light intensity (and thus X-ray intensity). A proportional voltage is produced and digitized to form the digital gray-scale image. Thus, scintillator-based TFT arrays, optically coupled CCD camera systems, and fiberoptically coupled CCD slot-scan systems as well as all PSP systems are classified as indirect acquisition devices.

Direct, in the context of this article (unlike the use of this term by many manufacturers that produce indirect acquisition DR devices), refers to the method of acquisition and conversion of absorbed X-ray energy into electron/hole pairs (charge) using semiconductor converters such as amorphous selenium (*a*-Se), solid-state silicon, or high-pressure gas. These direct acquisition detectors have a voltage applied to electrodes on opposite surfaces of the absorber/semiconductor material to separate and collect the generated electrons and holes. A small amount of energy (on the order of 50 eV) is needed to produce an ion pair for *a*-Se; for a 50-keV absorbed X-ray, hundreds of charge information carriers are generated. Similar amplification of charge per absorbed X-ray photon occurs in other direct converters as well. Electric field lines direct the ion pairs to the collection electrodes without lateral spreading, thus providing high intrinsic resolution by accurately mapping the X-ray photon absorption event to the corresponding del position.

Photon-counting detectors are currently configured in slot-scan geometry, composed of either high-pressure gas or solid-state silicon. These detectors measure absorbed X-ray photon events individually as counts instead of energy integration like all other detectors.[4] Since a count will be tallied independent of the photon energy, a signal-to-noise ratio advantage of up to 40% is achieved for the same number of X-ray photons absorbed in the detector compared with energy integration detectors, as there is no bias toward higher energy photons, and elimination of other noise sources accompanying an energy-integrator detector is possible. A limitation is the maximum count rate that can be sustained.

So, what is the "best" digital radiography detector? There is no straightforward answer, as advantages and disadvantages are based upon multiple comparisons of the following characteristics: detective quantum efficiency (DQE), spatial resolution, contrast resolution, dose efficiency, acquisition and display speed, radiographic positioning flexibility, image quality, ability to use existing radiographic equipment, integration with information systems and electronic networks, system costs, service costs, detector longevity, and survival in a hostile environment (eg, trauma imaging), among many other benchmarks. Matching an appropriate digital radiography device with its intended clinical purpose is an exercise in determining actual needs and patient throughput requirements prior to evaluating the type of DR system that will best meet those needs in a cost-effective and efficient manner.[5,6]

Conclusion

The terms *CR* and *DR* will continue to be used, but collectively we must think beyond the traditional CR versus DR comparisons, past the issues of X-ray converter materials, to clinical relevance, cost-effectiveness, dose efficiency, image processing functionality, overall image quality, proper use of digital radiography attributes (eg, variable speed characteristics and dose index values), quality-control phantoms and automated computer routines to verify proper function, patient throughput, uptime, reliability, longevity, service, and optimization in the clinical arena. Insisting on DR systems that can self-monitor and verify optimal performance through the automated analysis of quality-control phantom images will move the industry toward providing these important tools.

Finally, while dose efficiency is important, it shouldn't be the overriding consideration in determining the type of imaging system that best meets a specific clinical need, as all DR systems that are approved by the U.S. Food and Drug Administration must demonstrate a reasonable image quality over a typical range of incident exposures for clinical procedures. More important is an understanding and use of exposure index values that describe the "effective speed" of the detector to ensure that overexposures that are otherwise difficult to discern do not become the standard of practice as a result of complacency and/or ignorance.

References

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