Imaging Characteristics of New Screen/Film Systems for Cephalometric Radiography

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The advent of rare-earth screen/film combinations has brought substantial x-ray dose reductions with minimal loss in imaging quality. The physical characteristics of these recording systems are described and their potential for use in routine cephalometric sudies is evaluated.

There is an increasing awareness in both the scientific community and the public sector of potential hazards in even such small doses of radiation as those commonly employed in diagnostic radiological procedures. The subject of this report is the new technological developments in screen/film systems for radiographic imaging which allow greatly reduced radiation exposures and/or improved image quality in diagnostic procedures.

These advances are dependent on two favorable characteristics of the new phosphors—increased absorption of the incident x-rays and a higher efficiency in the conversion of that x-ray energy to light. That light covers a broader spectral range than is emitted by conventional Calcium tungstate phosphor, requiring orthochromatic films sensitive to the entire spectral

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output for maximum efficiency (speed).

The imaging properties of these new screen/film systems as they relate to cephalometric radiography are analyzed in this report. They are compared with both conventional calcium tungstate screens and the xeroradiographic modality.

Physical Characteristics

The x-ray film is the final detector and recorder of the image in a radiographic examination. The film may be exposed directly to the x-ray image emerging from the patient (a very inefficient method) or it may be sandwiched between two luminescent intensifying screens. These screens are composed of a phosphor of high atomic number which absorbs a larger fraction of the incident x-rays than the film, and emits a relatively bright corresponding image in visible light. It is that visible image which is recorded on the photographic emulsions of the film.

Until recently, calcium tungstate ($CaWO_4$) has been the principal phosphor used for x-ray imaging. Most orthodontists are currently using screens of this type. In the early 1970's, development began on application of new phosphors composed of terbium-activated rare-earth oxysulfides. These have some definite advantages over the conventional ($CaWO_4$) screen,[1] in that their higher absorption and conversion efficiency more efficiently utilizes the available quanta of x-ray energy.

Since exposure in diagnostic radiography must be minimized, it follows that the imaging system for a given radiographic procedure must be chosen so that exposure will be the lowest that will produce the required diagnostic information.

The performance of the new phosphors may be objectively quantified in terms of a set of well-defined parameters: system sensitivity (speed), contrast, latitude, resolution and noise.[2] A complex interplay of all of these parameters defines the final radiographic image quality.

We will first briefly describe these basic characteristics, and then apply them to cephalometry.

System Speed

The speed of a radiographic screen/film combination is a function of the combined properties of screens and film. It is defined much like the speed of photographic film in conventional photography, as the reciprocal of the exposure in Roentgens (R) or milli-Roentgens (mR) required to produce a film density* of 1.0 above base plus fog.

Screen speed is a function of two basic properties, efficiency and amplification. Efficiency is the fraction of incident X-ray photons absorbed and converted to visible light. Amplification defines the number of visible photons produced by the phosphor from each X-ray quantum absorbed.

Screen thickness strongly affects the screen speed because thicker screens absorb a higher percentage of the incident x-ray flux. The increased thickness, however, results in greater light diffusion and loss of resolution. The phosphor grain size and distribution also affect the speed through their effects on the light output of the screen.

* Density is a logarithmic measure of blackness. At a density of 1, 1/10 of the incident light is transmitted—at a density of 2, .01 of the incident light is transmitted—and at a density of 3, only .001 of the light is transmitted. Depending on lighting conditions densities greater than 2 to 2.5 will appear black.

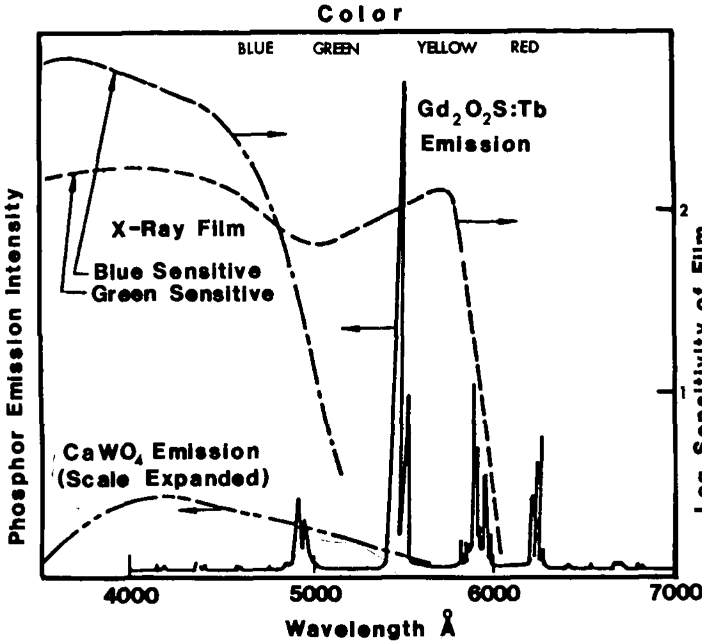

Fig. 1 Screen spectral emission and film spectral sensitivity for a conventional and a new screen/film system. Note that green-sensitive film is required to utilize the entire spectrum of visible light emitted by the rare-earth screen. *From Buchanan.*[1]

With respect to the film, the determining factor in speed is the silver halide grain size. X-ray film is actually a specialized black-and-white photographic film, so all the rules of photography apply.

The silver halide crystals can be produced in many different shapes and sizes, depending on how they are grown. In general, grains with large cross-sectional areas produce emulsions with higher speed but lower resolution than do small grains. Emulsion thickness is also a factor, working much the same as screen thickness.

An important consideration in achieving an optimum system sensitivity is that the spectral sensitivity of the film match the light emission spectrum of the screen. Different screen phosphors produce light of different colors, and films are made to be most sensitive to the colors emitted by specific screens.

Since the calcium tungstate intensifying screens emit in the blue range, the classical silver bromide emulsion which absorbs up to about 500nm has been satisfactory. With the advent of the rare earth phosphors such as Gadolinium, which also fluoresces in the green, it is necessary to use an orthochromatic film with spectral sensitivity extended into the longer wavelengths.

The spectral emissions of $CaWO_4$ and Gd_2O_2S:Tb are shown in Fig. 1, along with the spectral sensitivity of standard blue-sensitive film and a newer extended-range green-sensitive film. Notice that maximum speed for the Gadolinium rare-earth phosphor can be achieved only when it is used in combination with the green-sensitive film.

Contrast and Latitude

Unlike system speed, these parameters are functions of the film alone. They are photographic properties identical to those used in specifications for camera film. The image is produced by variation in exposure (light) across the area of the film. The information is carried by incident exposure variation ΔE, which is converted by the film into density variations ΔD which modulate the film density level.

The relationship between ΔE and ΔD is graphically described in the characteristic curve known to photographers as the H & D curve. This is a plot of density as the ordinate versus the logarithm of the exposure as the abscissa (Fig. 2). Note that exposure in this context is the actual light reaching the film at a specific point, which is only indirectly related to the exposure parameters shown on the x-ray control panel.

The linear portion of the H & D curve is mathematically represented by the following equation.

$$D = \gamma \log E + k, \qquad (1)$$

where γ is the straight-line slope, and k is a constant. By taking differentials it is very easy to show that

$$\Delta D = \frac{\gamma}{2.3} \cdot \frac{\Delta E}{E}, \qquad (2)$$

where E is the average exposure level.

Contrast

Equation 2 demonstrates how the radiographic contrast ΔD, which distinguishes anatomic structures of different density, depends on the subject contrast $\Delta E/E$ and the film *gamma.* Gamma is the numerical expression of film contrast, usually quoted as the slope of the H & D characteristic curve at a net density of 1.0.

Another closely related term is the *average gradient,* which covers the most useful part of the density range. This is the slope of the straight line drawn on the characteristic curve between the points on the curve at densities of 0.25 and 2.0 above the baseline density, which is made up of the combined densities of the plastic film base and emulsion fog.

The mathematical expression of average gradient is shown in the following formula, where E_1 and E_2 are the exposures required to produce net film densities of 0.25 and 2.0 respectively.

$$\textit{Average Gradient} = \frac{1.75}{\log E_2 - \log E_1} \qquad (3)$$

Latitude

Film latitude refers to the interval of exposure on the H & D curve which brackets the useful density range of the film. Latitude is often defined as the exposure range in logarithmic units to bring about that density change.

$$\textit{Latitude} = \log E_2 - \log E_1 \qquad (4)$$

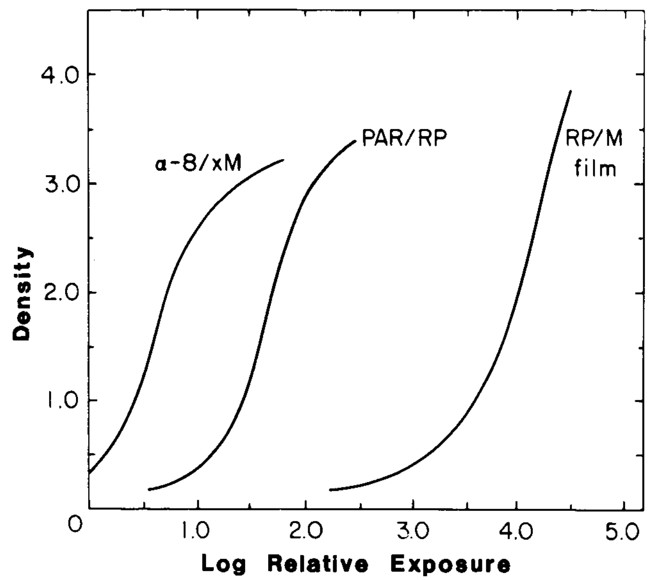

Fig. 2 Characteristic curves for three imaging systems. See text. Data from Gray.[2]

Contrast vs. latitude

It is clear from equations 3 and 4 that the average gradient and the latitude are inversely related. A high-contrast film provides excellent visualization of faint details, but over a more restricted x-ray intensity range. On the other hand, a film with a wide latitude allows a greater range of body tissues to be simultaneously displayed at the cost of low contrast and poor visualization of many structures such as tooth roots.

Both film latitude and average gradient are subject to considerable variations, depending on the conditions of development (temperature, time and developer formulation), as well as screen/film system and exposure.

In Fig. 2, the characteristic curves for three widely different systems are shown: a new high-speed rare-earth screen/film system (3M Trimax-8/XM), a conventional $CaWO_4$ system (Par/RP), and a direct exposure x-ray film (Kodak RP/M). The relative speeds and average gradients for the three systems are shown in Table 1.

Table 1

Relative speeds
and
Average gradients

System	Relative Speed	Average Gradient
Par/RP	100	2.6
Trimax-8/XM	800	2.5
RP/M	0.7	1.8

Resolution

The visibility of image details is affected by several factors. These include geometric blurring due to the size of the x-ray tube focal spot, subject contrast, x-ray scatter, light diffusion in the screens and film, and noise (or radiographic mottle). We will be concerned here only with the light diffusion and noise, which are directly related to the screen/film system.

Light Diffusion

Since light diffusion arises in the screens, while the resolving power of radiographic films is generally much greater, resolution of a screen/film system is heavily dependent on the screens.

When an x-ray photon is absorbed in an intensifying screen, several hundred photons of light are emitted randomly in all directions, and they undergo multiple scattering from nearby screen particles before they reach the film emulsion. As a result, an x-ray beam in the form of a narrow line will produce a "defocussed" image of a broader line.

For any given screen material, a thicker screen will cause a greater light spread, with a correspondingly greater loss of resolution. The thicker screen will, however, absorb a higher percentage of the incident x-ray flux; resolution and speed bear an *inverse* relationship to each other.[3]

If high resolution is required, as in detecting hairline fractures, a slow screen/film system or even direct-exposure film alone should be used. The resolution characteristics of directly-exposed film is far superior to any screen/film system, but at the cost of radiation levels that are excessive in any situation where such detail is not required. In cephalometry, where the emphasis is on the detection of high-contrast anatomical landmarks, reduced resolution in favor of higher speed should be reasonable.

The new generation of screens offers greater speeds than calcium tungstate, at comparable resolution, because of their (1) increased x-ray absorption for a given screen thickness and (2) increased light emission per unit of absorbed x-ray energy.

Modulation Transfer Function (MTF)

The information in an x-ray picture is presented on the film as a modulation of densities, from black to clear, in a pattern that represents the modulations imposed on the x-ray beam by the subject being radiographed. That modulated x-ray beam illuminates the screen/film sandwich in the cassette, and the effectiveness of the screen/film system in transferring those modulations of the x-ray beam to the permanent image on the film is objectively represented by the modulation transfer function (MTF).

That number is based on the reproduction of an x-ray beam that has been modulated into a sinusoidal pattern. If that pattern were reproduced perfectly on the film, it would be a series of stripes shading repeatedly through clear-gray-black-gray-clear, with a plot of those fluctuations in film density perfectly matching the sinusoidal plot of the x-ray beam fluctuations. Such a perfect reproduction would be represented by an MTF of 1, indicating that all of the information (modulation) was transferred to the film.

Such perfection is not attainable in practice. In practical application, the MTF lies somewhere between 0 and 1. Diffusion of the fluorescent light produced in the intensifying screens obliterates detail in the film by spreading light from the bright areas into the dark areas and wiping out some of the difference. The amplitude of the reproduction of the sine wave is diminished by the diffused light. This effect is smallest when the fluctuations are wide (low spatial frequency), because the diffused light does not spread all the way across the interval. The MTF reaches 1 when the x-ray beam is perfectly uniform, with no fluctuations (modulations) to transfer. As the fluctuations of the waves are adjusted closer together (high spatial frequency), the effect increases until at very high frequencies it all blurs into a uniform gray. At that extreme, with no transfer of the beam modulations, the MTF is 0.

What we have is a system in which the MTF begins at 1 for a frequency of 0 and drops as we introduce the fluctuations and move them closer together. In that midrange we see the original pattern on the film. As we continue to bring the lines closer together they become less clear and the MTF drops until it reaches 0 when we can no longer distinguish the pattern.

At each spatial frequency (usually expressed in cycles per mm) the fraction of input modulation which will survive in the image produced by the screen/film system is specified by the MTF. In general, the MTF falls rapidly from 1.0 at zero cycles/mm to 0.1 (or 10% transmission of signal amplitude) at 3 to 10 cycles/mm, depending on the screen/film system used.

Although the resolution characteristics of x-ray intensifying screens are best specified by a complete MTF curve covering the full range of frequencies, it is common practice to quote only one or two points on the curve. One common method is to present the spatial frequencies at which the MTF becomes 0.5 and 0.1.

The modulation transfer functions for three conventional calcium tungstate screens and two new rare-earth screens are shown in Figs. 3 and 4. The relative speeds of the Detail, Par and Hi-Plus systems are 1, 4, and 8 respectively. Notice that the slowest system (Detail) has the best resolution. For example, at two cycles/mm, the Detail system records 56% of the incoming sinusoidal signal, compared to only 29% for Hi-Plus. The signal transmission falls rapidly with increasing frequency for all systems. The Trimax-4 system has a resolution comparable to the Par system (Fig. 4), but it is considerably faster (Tables 2 and 3).

Equivalent Passband

A single-number measure of image sharpness is the equivalent passband, N_e, defined as the integral of the squared MTF:

$$N_e = \int_{-\infty}^{\infty} |MTF(f)|^2 \, df, \qquad (5)$$

where f is the spatial frequency

This index, introduced by Otto Schade,[4] correlates well with visual impressions of sharpness; N_e is directly proportional to the resolution characteristics of the imaging system, providing a simple index for comparing films and screens. The greater the value of N_e, the greater the resolving power (sharpness) of the system.

Noise

Noise is broadly defined as any unwanted disturbance. We are most familiar with its application to sounds as it affects our hearing, but it is also applied to electrical and visual dis-

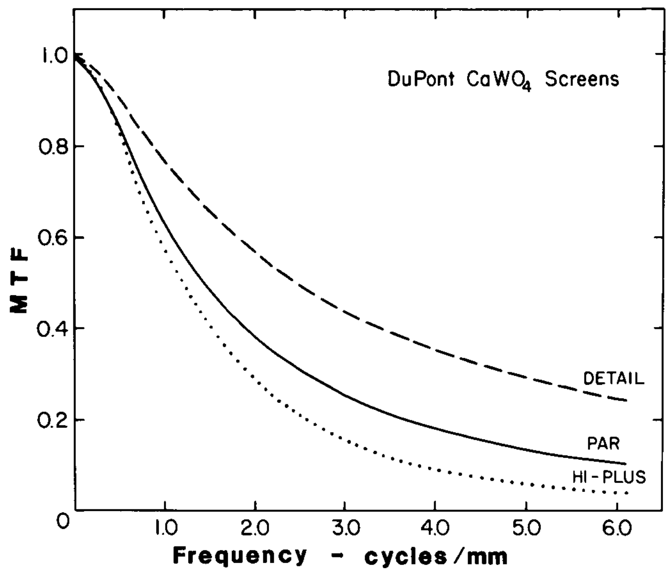

Fig. 3 The modulation transfer functions for three calcium tungstate screens. Notice the inverse relationship between resolution and speed. Data supplied by manufacturer.

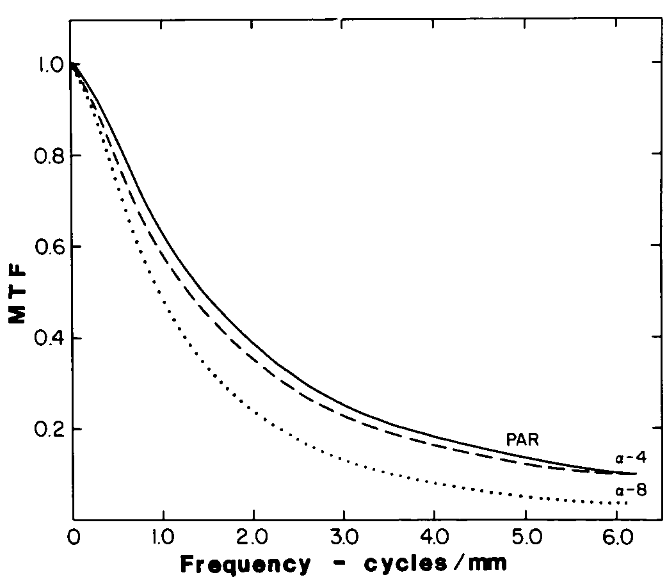

Fig. 4 The modulation transfer functions for two new rare-earth screens. Data supplied by manufacturer.

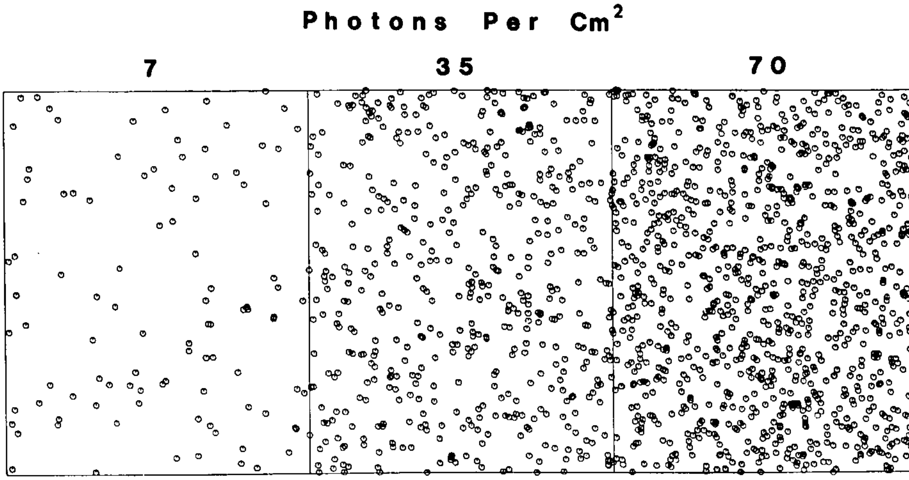

Fig. 5 The concept of quantum mottle: statistical fluctuations are more pronounced for low photon counts per unit area.

turbances. In visual imaging, including x-ray imaging, it refers to random fluctuations that appear in the image from sources other than the object being imaged.

When a film sandwiched between a pair of intensifying screens is exposed to a uniform x-ray beam, the resulting image is not uniform. The density varies from point to point, largely as a result of random spatial fluctuations in the incoming "uniform" x-ray beam, which is not truly uniform. An x-ray beam is made up of randomly distributed quanta of energy that form a more-or-less uniform pattern similar to that of a short burst of spray paint.

The statistical fluctuations in the number of x-ray quanta (N) incident over a given area in the film (the quantum mottle), are described by Poisson statistics with a standard deviation equal to $N^{1/2}$. The relative fluctuation is then $N^{1/2}/N = 1/N^{1/2}$. It is clear that the statistically predicted precision can be improved by using a large N, meaning a high x-ray exposure and a slow screen/film system.

The concept of quantum mottle is illustrated in Fig. 5. An image free of quantum mottle requires that the fluence (number of photons per unit area in the beam) be fairly large in order to minimize the statistical fluctuations. The faster rare-earth screen/film systems require such small x-ray exposures that quantum mottle can become a factor in detail visualization. It is like spraying too lightly to get a full, even coat of paint. This does not appear to be a problem in visualizing the large, high-contrast details used in cephalometry.

Overall mottle in an exposed film can be measured in terms of the Selwyn noise index. This is defined as the quantity $A\sigma^2_D$ where σ_D is the standard deviation of a number of density measurements on the film and A is the area of the aperture used in making the measurements. This is a gross index that does not reveal the source of the noise—how much is due to the spatial frequency of the x-ray beam and how much is due to the MTF of the screens or film grain.

It should be emphasized that image sharpness such as that described by the MTF affects the visualization of the input x-ray beam noise in the same way that it affects input diagnostic information. Sharp detail in the screen/film system may show the beam noise while a less sharp system will blur it out. Different tradeoffs may be engineered between sharpness and noise in a manner similar to the speed-sharpness tradeoffs discussed earlier.

A more sophisticated method of noise quantification involves the conversion of the density fluctuations on a uniformly exposed film into their Fourier components. The square of the Fourier transform of the measured density fluctuations, referred to as the Wiener spectrum, provides a complete description of the spectral composition of noise. A generalized Wiener spectrum of a typical screen/film system is shown in Fig. 6.

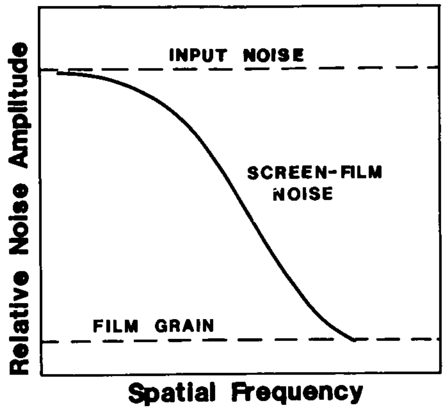

Fig. 6 The noise spectrum, or Wiener spectrum, of the screen/film system. The noise is high at low spatial frequencies (where the MTF is high), and rapidly falls with increasing frequency.

The input constant noise is modified by the system sharpness (MTF) and appears as the curve labeled Screen/Film Noise. The noise is high at low spatial frequencies and falls off gradually with increasing spatial frequency as detail is lost due to degradation of the MTF, eventually reaching the level of the film granularity. Film grain contributes a small component independent of frequency.

In practice, the Wiener spectrum presents the same problem as the MTF; it is a function rather than a number. The integral of the Wiener spectrum above the film granularity level is a measure of the total quantum mottle. This integral is proportional to $N_e(S/\varepsilon)^{1/2}$ where N_e is the equivalent passband, S the speed of the screen/film combination and ε the x-ray screen absorption factor, or efficiency.[5]

The quantum mottle index (QMI), $N_e(S/\varepsilon)^{1/2}$, provides a single-number representation of the total quantum mottle noise and illustrates mathematically the dependence of noise on the system's resolution (N_e), speed (S) and screen absorption (ε).

Quantum mottle increases with resolution, since a sharper system will transmit any input noise more faithfully to the film. A faster system will also be noisier because it will require fewer x-ray quanta to yield a given image density.

Additional graininess arises from the random distribution of phosphor particles in the screens and exposed grains in the film emulsion. Overall noise is a function of both film and screen.

Xeroradiography

Xeroradiography is another imaging modality[6] that has occasionally been suggested for cephalometric studies. It is an electrostatic technique. As the name implies, it is a completely dry process in contrast to the *wet chemistry* process of film development. A xeroradiographic system is exemplified by the Xerox 125 System, which consists of two units, the conditioner and the processor. The process is completely dry, and requires no dark room as all processing takes place within the two light-tight units. The basic element of xeroradiography is a thin layer (approximately 130 μm) of a photoconducting material (amorphous selenium) coated on a metal substrate.

The photoconductor is *conditioned* immediately before exposure with a uniform positive electrical charge by corona charging. Selenium is an excellent insulator in the dark, and the coating will hold the priming charge for a sufficient time to make an exposure. The sensitized plate is loaded automatically into a light-tight cassette.

The x-ray exposure is made in the same way as with a conventional screen/film system. X-rays absorbed in the selenium layer make it locally conductive, altering the uniform surface charge distribution into a latent electrostatic image.

The sealed cassette with the exposed plate inside is immediately inserted into the processor, where the plate is automatically removed from the cassette and transported to the development chamber. There, the image is made visible by spraying the plate with a cloud of charged colored particles which adhere mostly in regions of high electrostatic field strength. That powder image is then made permanent by transferring and fixing it on a special white paper.

The Selenium plate is subsequently cleaned and readied for a new charge and another x-ray exposure. A xeroradiographic image differs in appearance, formation and information content from a conventional silver halide image. It exhibits very limited macroresponse (the H & D curve is essentially flat), with detail detection dependent principally on border exaggeration. The subdued response leads to a wide recording latitude which permits the simulaneous display of many body structures of very different radiopacities.

Edge enhancement is seen as toner accumulation adjacent to a depletion zone. This reordering of the toner near edges makes xeroradiography a contouring process with resolution characteristics resembling those of the eye—there is a sharp response peak at about one cycle/mm. The lower frequency response (large areas) which carries less information is supressed.

A major drawback of xeroradiography is its relative insensitivity, which requires *much higher* x-ray exposures than screen/film systems. This has confined its use in medicine to the imaging of extremities.

Method and Materials

For appropriate evaluation of imaging systems for cephalometric radiography, the task must be defined in terms of objective imaging parameters.

Most structures to be imaged are large, with low spatial frequency. This means that high resolution (N_e) is not required. There is also high subject contrast (hard tissue), with great variation in density. Wide latitude is required for the broad density range, but the reciprocal relationship between latitude and contrast is not a serious problem with high-contrast structures.

Some exceptions are the imaging of tooth roots which requires high contrast, and imaging the soft tissue profile, which requires wide latitude.

Soft tissue profile imaging can be achieved by use of a wedge filter placed between the tubehead and subject, application if a contrast agent such as barium paste in a vertical stripe along the midline of the face, or special two-film techniques. All of these imaging requirements and techniques are compatible with use of a fast screen/film system to minimize patient exposure.

Several imaging systems were evaluated for use in cephalometrics. It must be pointed out that these systems represent only a sample of those currently marketed, being limited to those available at the authors' institutions. Other essentially equivalent systems are available from other manufacturers.

Changes in nomenclature of some of these systems have occurred recently. For example, the 3M screens formerly called Alpha-4 and Alpha-8 are now called Trimax-4 and Trimax-8 by the manufacturer, and faster Trimax-12 screens have also become available (the numbers indicate relative speed). Other nomenclature changes may occur, and users are cautioned against being misled by mere name changes.

Findings

Imaging characteristics of several suitable systems are summarized in Tables 2 and 3.[7] These data are summarized from reference 7, which provides more extensive tables listing the basic imaging characteristics for a number of radiographic screen/film systems.

To illustrate the use of these tables, consider a common screen/film combination, DuPont Par Screens/Cronex 4 film. From Table 3 we find that the speed (S) of this system is 0.8 mR^{-1}. This implies that under conditions of narrow-beam geometry and heavy filtration, the exposure required at the screen/film position to achieve a net density of 1 is 1/0.8, or 1.25 mR at 80 kVp. The overall resolution of this combination is indicated by its N_e value which is 2.3. The quantum mottle index is 4.5 and the gamma 3.0.

If we now look at the Trimax-8/XM screen/film combination, we find from Table 2 that S equals 8 mR^{-1}, so the required exposure is 1/8 or 0.13 mR. Since this system is eight times faster than the Par/Cronex 4 system, it requires only ⅛ of the exposure. Its higher speed, however has also resulted in somewhat lower resolution ($N_e = 1.8$) and higher noise (QMI = 6.6).

Test exposures using several screen/film combinations were made using standard radiographic equipment and a tissue-equivalent head mannequin. The distance from the x-ray source to the midline of the head mannequin was 60 inches, as commonly employed in cephalometric radiography. The cassette containing the film and screen was placed as close as possible to the exit surface of the mannequin. The resulting radiographs are shown in Figs. 7-11.

Image characteristics of all of these radiographs are suitable for cephalometric analysis as typically used. There are differences in esthetic qualities, but the information required for landmark identification is present in all. For the radiographs made with screen/film systems (Figs. 7-9), the system contrast, latitude and resolution are about the same; major differences are in speed and noise.

Table 2

Rare-earth green-emitting screens and green-sensitive films speed/quantum mottle index

	Film	Screen ►	Kodak Lanex Fine Ne 3.2	Kodak Lanex Regular Ne 1.8	3M Trimax-4 Ne 2.0	3M Trimax-8 Ne 1.6
Kodak	Ortho-G	γ2.8	1.0/5.5	3.5/4.3	—	—
	Ortho-L	γ2.2	1.0/5.5	3.5/4.3	—	—
3M	XD	γ2.9	—	—	2.0/4.4	4.0/4.6
	XM	γ2.5	—	—	4.0/6.2	8.0/6.6

Table 3

Conventional blue-emitting screens and blue-sensitive films speed/quantum mottle index

	Film	Screen ►	DuPont Par Ne 2.3	DuPont Hi-Plus Ne 1.9	DuPont Quanta ii Ne 1.8	Kodak X-omatic Regular Ne 2.2
Kodak DuPont	XG Cronex 7	γ3.0 γ3.0	0.4/3.2	0.8/2.9	1.6/3.7	0.8/3.2
Kodak DuPont DuPont DuPont 3M	XRP Cronex 4 Cronex 6 Cronex 6+ Type R	γ2.8 γ3.0 γ2.2 γ2.6 γ2.4	0.8/4.5	1.6/4.1	3.2/5.2	1.6/4.6

Ne is the equivalent passband
The paired values are speed/quantum mottle index as defined in the text.

Fig. 7 Simulated cephalometric radiograph using high-speed conventional screen/film system (Dupont Hi-Plus screens, Cronex 6+ film), with 4:1 grid.

Figs. 7 and 8, made with the same screen/film system without and with a 4:1 parallel grid, demonstrate the effect of scatter from within the subject on image noise. Clearly, use of the grid results in greater visibility of small details, but at the penalty of considerably higher exposure (Table 4).

As discussed previously, latitude is greatest in the xeroradiographic image (Fig. 10). Direct-exposure film (Fig. 11) provides greater latitude and resolution than screen/film systems, but with reduced contrast and sensitivity.

Radiation Risks

Risks to patients from small doses of radiation are cancer and mutation.[8] They are stochastic in nature; the probability of effect in a given individual is a function of dose. The probability of these effects from conventional intraoral and panoramic radiography has been recently estimated by Bengtsson,[9] and Danforth and Gibbs.[10]

The organs known to be at risk of radiation-induced cancer in the head and neck are thyroid, bone marrow

Fig. 8 Same as Fig. 7, without grid. Note increased noise, but film is still suitable for cephalometric analysis and exposure has been greatly reduced.

(leukemia), salivary glands and brain. For genetic effects, the gonads carry the risk.

For intraoral and panoramic radiography, reasonably good estimates of doses to these organs are available in the literature,[10] allowing for numeric estimation of patient risk. For cephalometric radiography, however, little published information on radiation doses to the organs can be found for the various image receptor systems examined in the present study. We have measured skin entry exposures (Table 4), but these are not reliable indicators of risk. Doses to internal organs depend not only on skin exposure, but also on beam energy (kVp and filtration) and beam size.

Doses to marrow and thyroid, calculated using the organ dose index system developed by Rosenstein,[11] are shown in Table 4. This system does not provide for calculation of doses to brain and salivary glands, whose sensitivity to radiation-induced cancer has only recently been identified.[8] Marrow doses in Table 4 are averaged

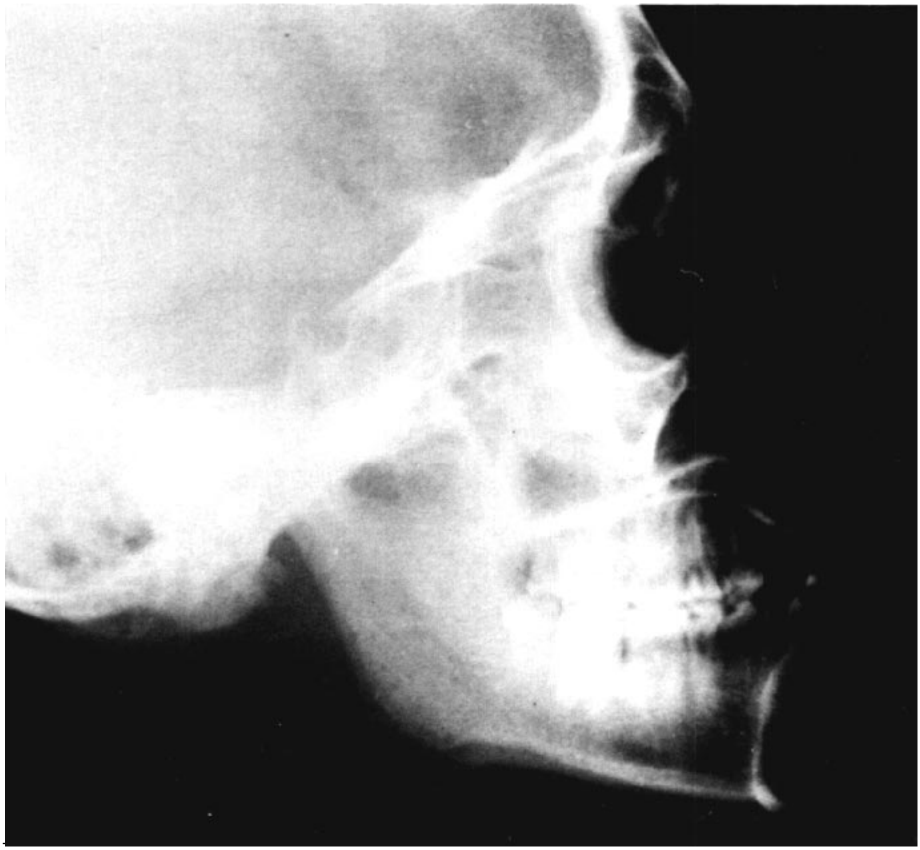

Fig. 9 Simulated cephalometric radiograph with high-speed rare-earth screen/film system (3M Trimax-8 screen, XM film). Patient exposure is significantly reduced, and image quality remains adequate. The difference between this image and Fig. 7 is increased noise; resolution and contrast are essentially unchanged.

over the entire marrow, using the method of Shleien,[12] since leukemia risks are expressed as probability of leukemia per unit dose to the entire marrow. Thyroid doses assume exclusion of the thyroid gland from the primary beam by collimation or by shielding. If the thyroid is exposed to the primary beam, doses will be substantially greater.

For comparative purposes, marrow doses from a full-mouth series of intraoral films are about 8-16 mrad (80-160 μGy), and from panoramic about 2 mrad (20 μGy). Thyroid doses are 10-70 mrad (100-700 μGy) from full-mouth intraoral, and 27 mrad (270 μGy) from panoramic examinations.

In the absence of reliable dosimetry data, only very crude estimates of patient cancer risk from cephalometric radiography can be provided. The following data must be regarded as little more than educated guesses, but are the best available.

Since all are based on the same un-

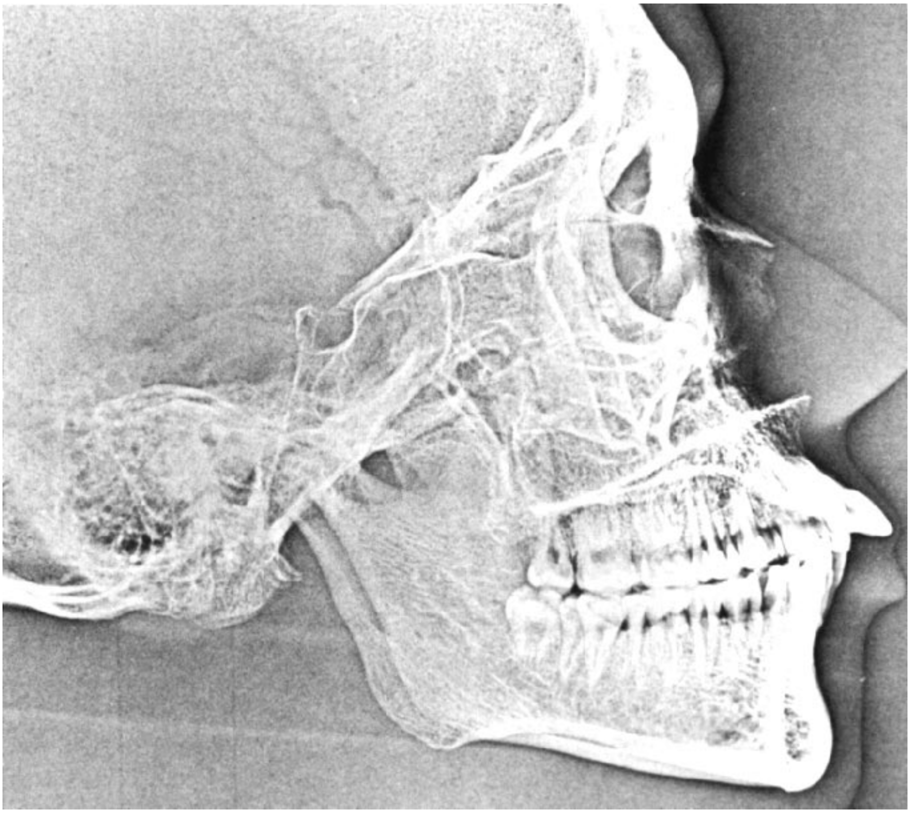

Fig. 10 Xerographic radiograph. Image quality is excellent, but patient dose is excessive.

derlying risk estimate the relative risks among different systems are valid even though absolute risks are merely crude estimates.

From conventional imaging (Hi-Plus screens, Cronex 6+ film, no grid), total cancer risk from a single lateral cephalometric exposure is of the order of 100 to 300 cases per billion examinations. Addition of a 4:1 grid would approximately double the risk. Xeroradiography would carry a risk about twenty times greater.

On the other hand, conversion to a fast rare-earth system (Trimax-8 screens, XM film) would reduce the risk by a factor of six, to about 16 to 50 cases per billion examinations.

Analysis of these risk approximations and of the films in Figs. 7-10 demonstrates the difficulty in justifying continued use of conventional imaging systems for cephalometrics.

Use of grids is even more difficult to justify; they do not contribute significantly to the information used for cephalometric analysis, yet at least double the patient risk.

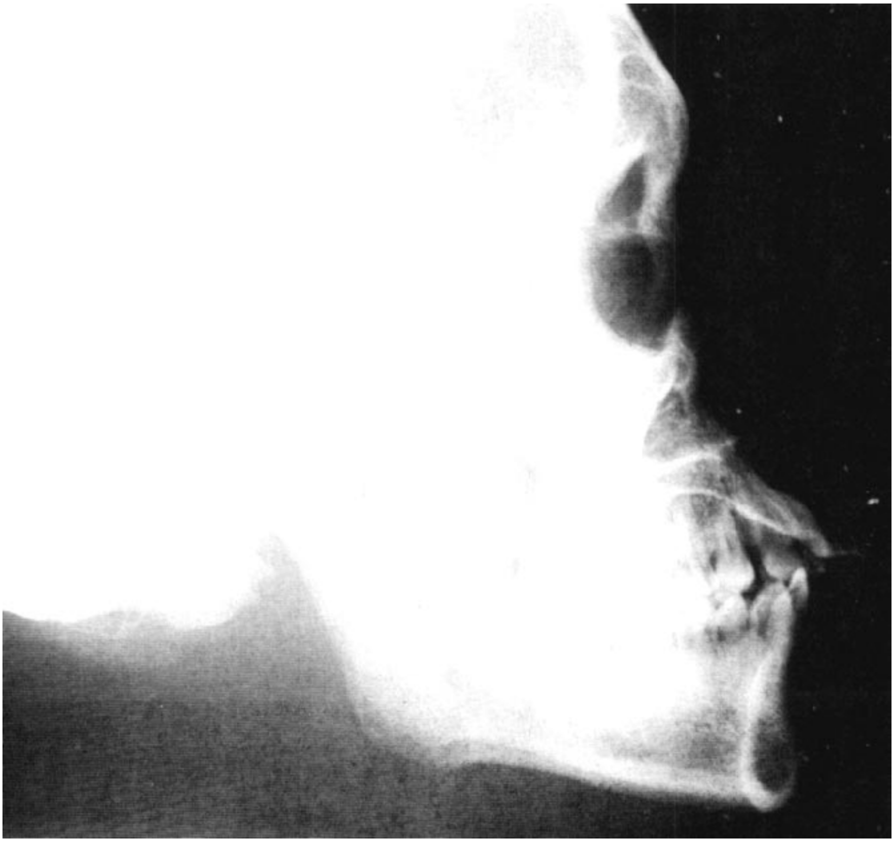

Fig. 11 Simulated cephalometric radiograph with no-screen film. Resolution and latitude are greater than with screen/film systems, but contrast is low and patient dose is excessive.

Xeroradiography, although providing esthetically attractive images, adds little in return for the additional risk, which is estimated as more than 100 times that of high-speed rare-earth imaging systems.

Conclusions

Several high-speed rare-earth screen/film combinations have been evaluated and compared with conventional imaging systems for cephalometric radiography. Results indicate that the rare-earth imaging systems can be utilized without significant loss of diagnostic information while allowing marked decreases in patient exposure. Therefore, these newer imaging systems should become the standard of cephalometric practice. A clinical evaluation of the new systems has been presented elsewhere.[13,14]

Table 4

Radiation dose from a lateral cephalometric radiograph

KVP	Screen	Film	Grid	Skin exposure mR	Marrow dose mrad	Thyroid dose mrad
70	None	Cronex 6+	4:1	2060	45	198
110	None	Xerox	None	125	5.9	20
70	Par	Cronex 6+	4:1	50	1.1	4.8
70	Hi-Plus	Cronex 6+	4:1	26	0.6	2.4
70	Hi-Plus	Cronex 6+	None	13	0.3	1.2
70	Trimax-8	XM	4:1	6	0.1	0.5
70	Trimax-8	XM	None	3	0.05	0.25

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