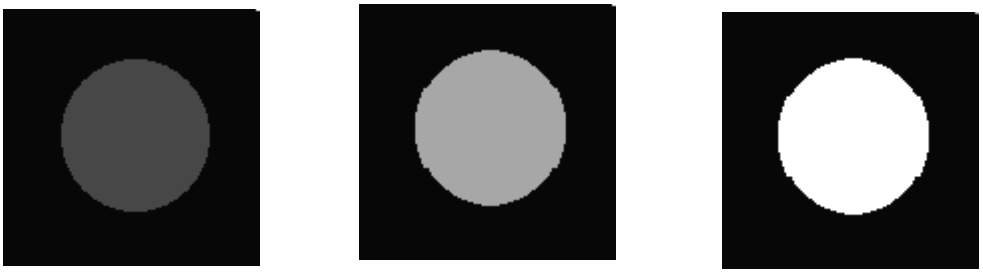
# CHAPTER 4: PHYSICAL DETERMINANTS OF CONTRAST

Any medical image can be described in terms of three basic concepts that we already have mentioned: contrast, spatial resolution or clarity, and noise. The spatial resolution or clarity refers to the spatial extent of small objects within the image. Noise refers to the precision with which the signal is received; a noisy image will have large fluctuations in the signal across a uniform object while a precise signal will have very small fluctuations. The subject of this discussion, image contrast, refers to the difference in brightness or darkness in the image between an area of interest and its surrounding background. For example, if gray and white dots are painted onto a black canvas, the white circle is presented with a larger contrast with respect to the background than the gray circle (Figure 4-1). The information in a medical image usually is presented in "shades of gray". (Color is avoided because it creates false borders that can distract the observer.) One uses the differences in gray shades to distinguish different tissue types, analyze anatomical relationships, and sometimes assess physiological function. The larger the difference in gray shades between two different tissue types, the easier it is to make important clinical distinctions. It is often the objective of an imaging system to maximize the contrast in the image for a particular object of interest, although this is not always true since there may be design compromises where noise and spatial resolution are also very important. The contrast in an image depends on both material characteristics of the object being imaged as well as properties of the device(s) used to image the object. In this chapter we will detail the concept of contrast and describe the physical determinants of contrast, including material properties, x-ray spectra, detector response, and the role of perturbations such as scatter radiation and image intensifier veiling glare.

**Figure 4-1**. Contrast



Low Contrast | Medium Contrast | High Contrast

## 4.1. Definition of Contrast and Physical Determinants of Contrast

In its most general terms, contrast can be defined as the fractional difference in some measurable quantity in two regions of an image. Usually when we say "contrast", we mean image contrast, which is the fractional difference in optical density or brightness between two adjacent regions in an image. In conventional radiography, image contrast depends on two other types of "contrast" called (a) radiographic contrast and (b) detector contrast. Radiographic contrast (sometimes called subject contrast) refers to the difference in the number of x-ray or gamma ray photons emerging from adjacent regions of the object being scanned, which depends on differences in atomic number, physical density, electron density, thickness, as well as the energy spectrum of the x-ray beam emitted by the source.

(Because radiographic contrast depends on the x-ray energy, which is not a characteristic of the subject, we avoid the use of the term "subject contrast" even though this the term commonly found in other texts.)

Detector contrast refers to the ability of the detector to convert differences in photon fluence across the x-ray beam into differences in optical density (film), image brightness (image intensifiers), signal amplitude (electronic detectors), or some other physical, optical, or electronic signal used to represent the image in the imaging system. As we shall see later, for many systems the measurement we make of the image signal must be linearly related to the intensity of the radiation signal generating the image. For some cases it will be necessary to use the H&D curve of the film to convert film densities to relative exposure values. The detector contrast depends on the chemical composition of the detector material, its thickness, atomic number, electron density, as well as the physical process by which the detector converts the radiation signal into an optical, photographic, or electronic signal. The detector contrast also depends on the x-ray spectrum used to image the object. The detector may increase or decrease the radiographic contrast; that is, the detector may produce photographic or electronic signals that have a larger or smaller fractional difference between adjacent areas of the image in comparison to the difference found in the radiographic signal.

A third component in the description of image contrast includes various physical pertubations such as scattered radiation, image intensifier veiling glare, and the base and fog density levels of film. Each of these tends to reduce image contrast. Also, digital image displays allow the observer to control window and level parameters that affect brightness and contrast. This provides a fourth component of contrast, which we will call "display contrast". Each of these components of contrast will be discussed in this chapter (Figures 4-2 and 4-3).

**Figure 4-2**. Components Of Image Contrast

**Radiographic (Subject) Contrast**
- Material Thickness
- Physical Density
- Electron Density
- Elemental Composition (Effective Z)
- X-Ray Photon Energy

**Detector Contrast**
- Detector Type (Film vs. Electronic)
- Film Characteristic Curve (H&D)
- Spatial Response of Detector

**Physical Perturbations**
- Scatter Radiation
- Image Intensifier Veiling Glare
- Base and Fog of Film

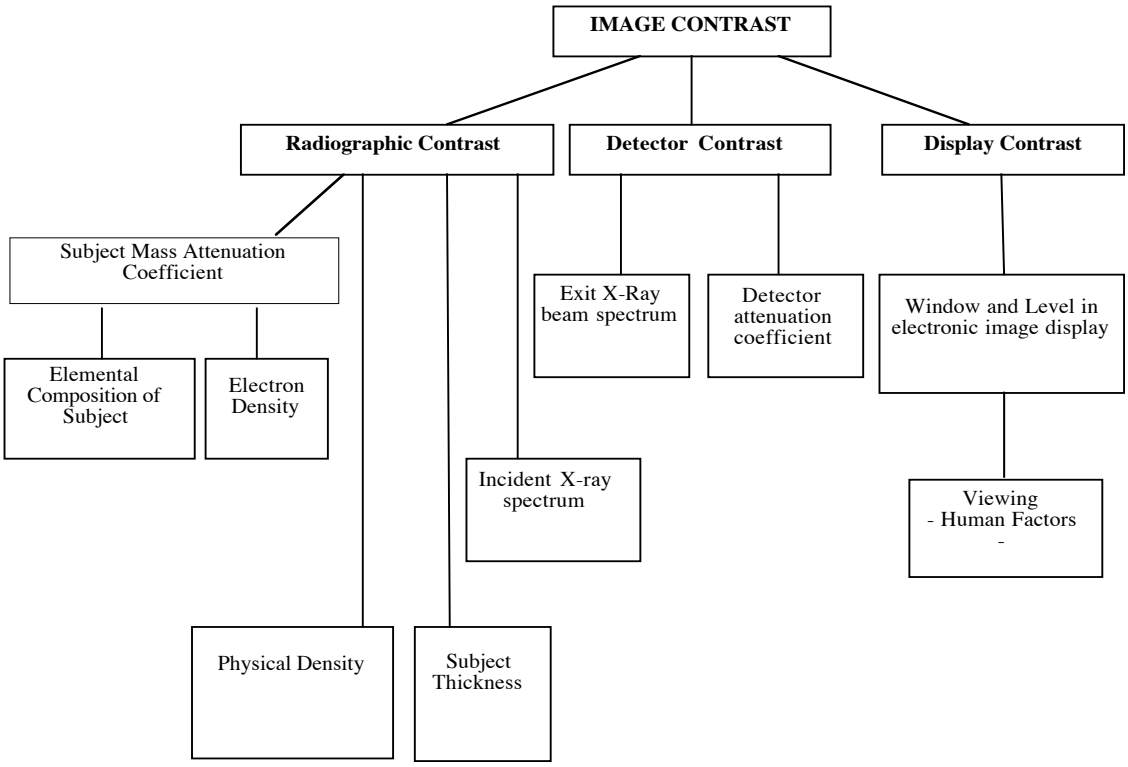
**Display Contrast**
- Window and Level Settings

Finally, the training and physiological status of the human viewer (e.g., myopia, astigmatism, color blindness, fatigue) as well as environmental factors such as high ambient light levels in the viewing area also affect how one perceives the contrast presented in the image. Training and/or memory of similar images also impacts visual perception of the information in a radiograph. However, in this book, we will not include these factors with the other components of contrast. Nevertheless, they are essential components of the imaging system and deserve special attention.

## 4.2 Radiographic Contrast

The physical determinants of contrast can be understood by examining the processes by which a radiographic image is formed. We will consider a system in which a patient is placed between an x-ray tube and a detector, the detector being either a film-screen combination or an electronic detector. The x-ray tube is operated at a certain kVp, which, along with any filtration, determines the energy spectrum of beam. X-ray photons from the source are attenuated by various materials in the patent (muscle, fat, bone, air, and contrast agents) along the path between the source and detector.

**Figure 4-3**. Components Of Image Contrast




The photon attenuation of each material depends on its elemental composition as well as the energy of the beam. This effect is assessed using its linear attenuation coefficient ($\mu$), which gives the fraction of photons absorbed by a unit thickness of the material. An equation like (4-1) is useful as mass attenuation coefficients are commonly tabulated rather than linear attenuation coefficients ($\mu_m$).

$\Phi_0$ = number of photons/area entering the volume
$\Phi$ = number of photons/area leaving the volume
$\rho$ = density of the material.
x = the thickness of the material
$\mu_m(E) = \frac{\mu(E)}{\rho}$ = mass attenuation coefficient as a function of the photon energy E, then

$$\Phi(E) = \Phi_0 e^{-\mu(E)x} = \Phi_0 e^{-(\mu(E)/\rho)\rho x} = \Phi_0 e^{-\mu_m(E)\rho x} \qquad (4\text{-}1)$$

The components of radiographic contrast are apparent in equation 4-1. The mass attenuation coefficient $\mu(E)/\rho$ depends explicitly on energy (E) and implicitly on the atomic number of the material and its

electron density. The second component ρ is the mass density of the material. The greater the density, the larger the attenuation afforded by that material, as seen in the product ρx. The third component x represents the thickness of the material. Again, the thicker the material the more attenuation that material provides to the x-ray beam. The energy of the x-ray beam determines the value of the mass attenuation coefficient and is one of the most important factors in controlling both the radiographic and overall (image) contrast. The mass attenuation coefficients generally decreases as the photon energy increases except at points of discontinuity called absorption edges, mostly the K-edge or L-edge. At these energies, photoelectric interactions between the photon and inner shell electrons cause large increases in the photoelectric cross-section as the photon energy slightly exceeds the binding energy of the orbital electrons. Correspondingly, contrast tends to decrease as the photon energy increases except when K-edge or L-edge discontinuities cause large increases in contrast. (As we will see, this is a consideration especially when contrast agents are utilized.)

If we could count photons on the detector-side of the patient, we could determine the radiographic contrast at this point in the image formation process (Figure 4-4). As given before, the radiographic contrast is defined to be the fractional difference in photon fluence between adjacent areas. For example, if behind the patient, photon fluence of $\varphi_1$ is measured in one area while a photon fluence of $\varphi_2$ is measured in an adjacent area (Figure 4-4), then the radiographic contrast is defined to be

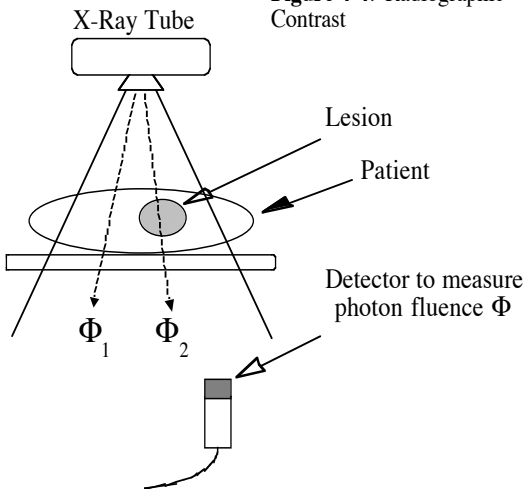



**Figure 4-4**. Radiographic Contrast

$$C(\Phi) = \frac{\Delta\Phi}{\Phi_1} = \frac{\Phi_2 - \Phi_1}{\Phi_1} \qquad (4\text{-}2)$$

(4-2)

For an opaque object, one where $\Phi_2 = 0$ then C= –1, though we usually just say that the contrast is 100%. When $\Phi_1 < \Phi_2$ radiographic contrast is positive and can exceed 100%. If $\Phi_1 = \Phi_2$, the object cannot be differentiated from its background (C = 0).

## 4.3 Radiographic Contrast of Biological Tissues

One of the principal determinants of contrast in a radiograph of the human body is, of course, the types of tissue found in the body region being imaged. The radiation attenuation properties of each tissue type in turn is determined by its elemental and chemical composition, as it is for all other (i.e. biologic and non-biologic) chemical compounds and mixtures. For purposes of our discussion, we can consider the body being composed of three different tissues: fat, (lean) soft tissue, and bone. In addition to these tissues, we must consider the attenuation properties of air found in the lungs (and in the gastrointestinal tract), as well as the properties of contrast agents that possibly will be introduced into the body.

The chemical composition of the three major tissue types are given in Table 4-1 and will be used in the description of attenuation properties presented in the following section. Some of their relevant physical properties are given in Table 4-2 and graphs of their mass attenuation coefficients as a function of energy are presented in Figure 4-5.

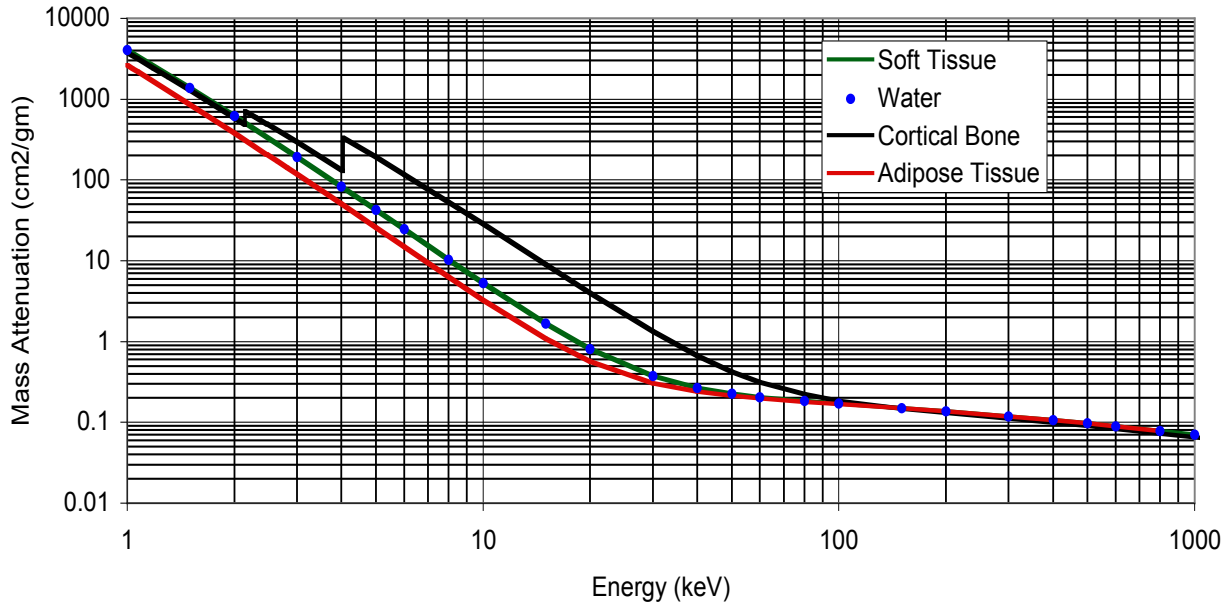
**Table 4-1**. Elemental Composition of Body Materials

| % Composition (by mass) | Adipose Tissue (fat) | Muscle (striated) (soft tissue) | Water | Bone (Femur) |
|---|---|---|---|---|
| Hydrogen (**Low Z**) | **11.2** | **10.2** | **11.2** | **8.4** |
| Carbon | **57.3** | **12.3** | | **27.6** |
| Nitrogen | 1.1 | 3.5 | | 2.7 |
| Oxygen | **30.3** | **72.9** | **88.8** | **41.0** |
| Sodium | | 0.08 | | |
| Magnesium | | 0.02 | | 0.2 |
| Phosphorus | | 0.2 | | 7.0 |
| Sulfur | 0.06 | 0.5 | | 0.2 |
| Potassium | | 0.3 | | |
| Calcium (**High Z**) | | 0.007 | | **14.7** |

**Table 4-2**. Physical Properties of Human Body Constituents

| Material | Effective Atomic No. | Density ($gm/cm^3$) | Electron Density (electrons/kg) | |
|---|---|---|---|---|
| Air | 7.6 | **0.00129** | $3.01 \times 10^{26}$ | lowest atten. |
| Water | 7.4 | 1.00 | $3.34 \times 10^{26}$ | |
| Soft tissue | 7.4 | 1.05 | $3.36 \times 10^{26}$ | |
| Fat | **5.9-6.3** | **0.91** | $3.34\text{-}3.48 \times 10^{26}$ | |
| Bone | **11.6-13.8** | **1.65-1.85** | $3.00\text{-}3.19 \times 10^{26}$ | highest atten. |

**Figure 4-5**. Mass Attenuation Coefficients of Tissues ($cm^2/gm$).




Soft Tissue

The term "soft tissue", as used in this text, excludes fat but includes muscle and body fluids. The term "lean soft tissue" sometimes is used to describe nonfatty soft tissues, but we will use the less cumbersome term "soft tissue" and implicitly exclude fat. There are, of course, many different types of soft tissue including liver tissue, collagen, ligaments, blood, cerebrospinal fluid, and so on. However, the chemical composition of these tissues is dominated by elements with low atomic numbers. Therefore, we will assume that they are radiographically equivalent to water and have an effective atomic number of 7.4 and an electron density of $3.34 \times 10^{26}$ electrons per gram. This assumption is plausible since, as Ter Pogossian has pointed out, soft tissues are approximately 75% water by weight while body fluids are 85% to nearly 100% water by weight.

Water-equivalent tissues have several important radiologic properties that contribute to their contrast. First, the photoelectric effect dominates photon attenuation up to energy of 30 keV, after which the Compton effect becomes increasingly dominant in the remainder of the diagnostic energy range. As we will see later, photoelectric interactions provide better differentiation between tissue types. Therefore it is desirable to use lower energy photons to maximize contrast in diagnostic examinations. Second, because the body is about 70% water-equivalent by weight, contrast for these tissues is dictated predominantly by variations in the thickness or density (i.e. by the product of $\rho x$ in equation 4-1). In the diagnostic range, the HVL of soft tissue is in the range of 3 to 4 cm so that thickness differences of 3cm provide a radiologic contrast of 50%. Finally, the radiographic similarity of a majority of tissue volume in the human body complicates our imaging task. For example, it is impossible to visualize blood directly or to separate tumors from surrounding normal soft tissue. This forces radiologists to use "contrast agents" (as will be discussed below) to provide contrast to enable visualization of these anatomic structures of interest. Without contrast agents, except in the grossest manner it is impossible to

visualize important structures, such as the liver, GI tract, or cardiac blood pool, using standard plane film x-ray techniques.

Fat

Along with the energy of the x-ray beam, electron density, physical density, and atomic number determine the attenuation of any material through their impact on the attenuation coefficient. Due to the presence of low atomic number elements, fat has a lower physical density and lower effective atomic number, and therefore a lower photoelectric attenuation coefficient, than either soft tissue or bone. For this reason, fat has a lower attenuation coefficient than other materials in the body (except air) at low energies where the photoelectric interactions are the dominant effect.

However, at higher energies, fat has a somewhat higher Compton mass attenuation coefficient than other tissues found in the body. Unlike other elements, the nucleus of hydrogen is free of neutrons, giving hydrogen a higher electron density (electrons/mass) than other elements. Because hydrogen contributes a larger proportion of the mass in fat than it does in soft tissue and bone, fats have a larger electron density than other tissues. This becomes particularly important at higher energies where Compton interactions dominate attenuation. In fact, inspection of a table of mass attenuation coefficients (Figure 4-5) shows that at higher energies the mass attenuation coefficient of fat slightly exceeds that of bone or soft tissue, precisely due to the higher electron density of fat. However, due to its low density it does not have a higher linear attenuation coefficient.

As Tables 4-1 and 4-2 show, the differences in atomic number, physical density, and electron density between soft tissue and fat are slight. The differences in the linear attenuation coefficients and therefore in radiographic contrast between fat and soft tissue is small. One must depend on the energy dependence of the photoelectric effect to produce contrast between these two materials. This is particularly true in mammography where one uses an x-ray beam with an effective energy of about 18 keV. Such a low energy spectrum maximizes the contrast between glandular tissue, connective tissue, skin, and fat, all of which have increasingly similar attenuation coefficients at higher photon energies.

Bone.

The mineral component of bone gives it excellent contrast properties for x-ray photons in the diagnostic range. This is due to two properties. First, its physical density is 60% to 80% higher than soft tissue. This increases the linear attenuation coefficient of bone by a proportionate fraction over that of soft tissue. Second, its effective atomic number (about 11.6) is significantly higher than that of soft tissue (about 7.4). Since the photoelectric mass attenuation coefficient varies with the cube of the atomic number, the photoelectric mass attenuation coefficient for bone is about $[11.6/7.4]^3 = 3.85$ times that of soft tissue. The combined effect of its greater physical density and its larger effective atomic number gives bone a photoelectric linear attenuation coefficient approximately 6 times greater than that of soft tissue or fat. This difference decreases at higher energies where the Compton effect becomes more dominant. However, even at higher energies, the higher density of bone still allows it to have excellent contrast with respect to both soft tissue and fat. Therefore when imaging bone, one can resort to higher energies to minimize patient exposure while maintaining reasonable contrast instead of resorting to low x-ray beam energies as one is compelled to do when attempting to differentiate fat from soft tissue.

## 4.4 Contrast Agents

Most of the methods we use to improve contrast involve control of variables external to the patient, such as detector response and choice of x-ray beam kVp. The factors that control contrast within the patient, such as the thickness, physical density, and elemental composition of the body's tissues, are difficult if not impossible to control while an image is being recorded.

There are times, however, when the composition of the body part can be modified to increase radiographic contrast. This is accomplished by introducing a material, called a contrast agent, into the body to increase, or sometimes decrease, the attenuation of an object being imaged (Figure 4-6). For example, agents containing iodine commonly are injected into the circulatory system when an angiographer is imaging blood vessels or the ventricular blood pool. This is necessary because blood and muscle both have attenuation coefficients essentially equal to that of water. Therefore, blood cannot be distinguished from surrounding soft tissue structures using conventional x-ray techniques without contrast agents. When an iodinated contrast agent is introduced into the circulatory system it increases the x-ray attenuation of the blood, allowing the radiologist to visualize the blood pool in the vessels (either arteries or veins), or in the cardiac chambers. Barium is another element that is commonly used as a contrast agent, particularly in the gastrointestinal tract. A thick solution containing barium is introduced into the gastrointestinal tract by swallowing or through some other path. When the barium solution is inside of the GI tract, the borders of the GI tract can be visualized so that the radiologist can look for ulcerations or any ruptures that may be present.

**Figure 4-6**. Examples Of Contrast Media

**Hydopaque (Iodine)**

Composition: 0.25 grams $C_{18}H_{26}I_3O_9$ + 0.50 grams $C_{11}H_{13}I_3N_2O_4$ + 0.6 grams water

Physical density: 1.35 grams/cm$^3$

K-edge Energy: 33.2 keV

Atomic Number of Iodine: 53
Applications: Angiography, Genitourinary (GU) Studies

**Barium sulfate ($BaSO_4$)**

Composition: 450 grams barium sulfate + 25 milliliters water

Physical density: 1.20 gm/cm$^3$

K-edge Energy: 37.4 keV

Atomic Number of Barium: 56
Applications: Gastrointestinal (GI) Studies

**Air**

Composition: 78% $N_2$ + 21% $O_2$

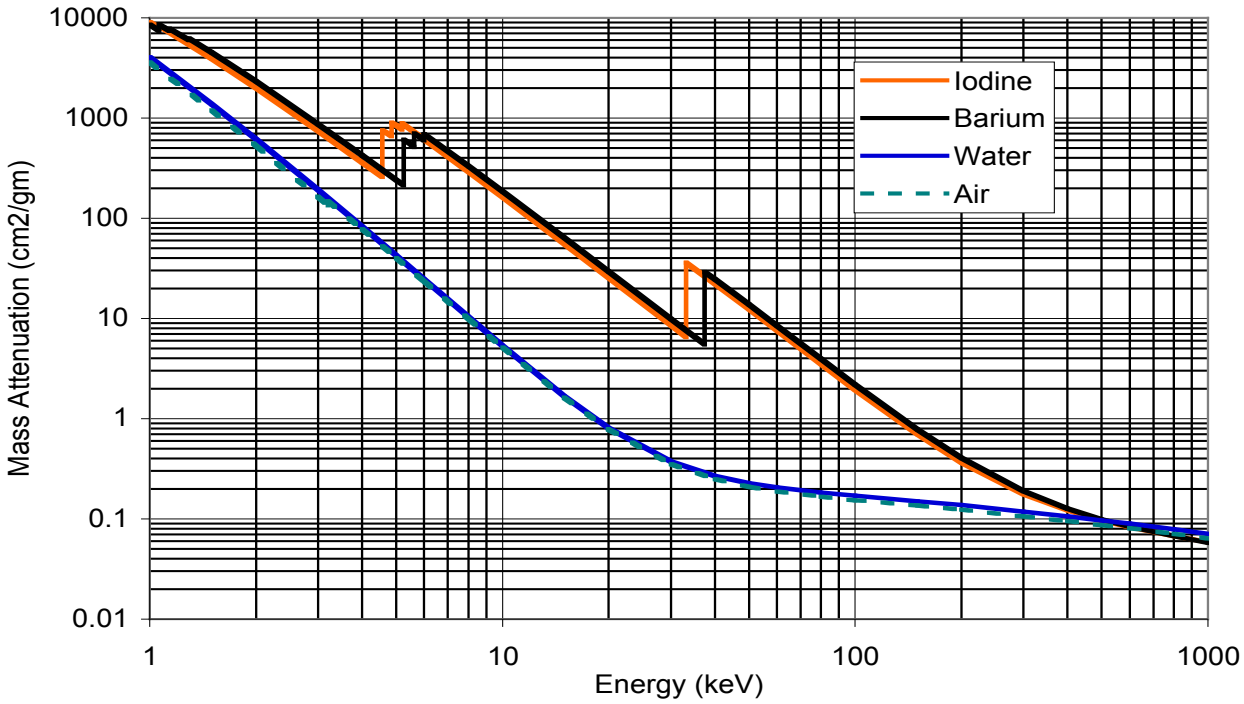
Physical Density: 0.0013 gm/cm$^3$

K-edge Energy: 0.4 keV

Effective Atomic Number: 7.4
Applications: GI Studies, Pneumoencepalography

Iodine and barium are used as contrast agents for several reasons. The first is that they can be incorporated into chemicals that are not toxic even when rather large quantities are introduced into the body. Second, to be useful as a contrast agent, the material must have a linear attenuation coefficient that is different from that of most materials found in the human body (Figure 4-7). When iodinated contrast agents are used in angiography, the iodine must provide sufficient x-ray attenuation to provide discernable contrast from surrounding soft tissues when imaged with x-rays.

**Figure 4-7**. Mass Attenuation Coefficients (cm2/gm) for Contrast Agents.




Both barium (Z = 56) and iodine (Z = 53) meet these requirements. A common iodinated contrast agent is Hydopaque (Figure 4-6), a cubic centimeter (cc) of which contains 0.25 grams of $C_{18}H_{26}I_30_9$, 0.50 grams of $C_{11}H_3I_3N_20_4$, and 0.6 grams of water with a density of 1.35 grams/cm$^3$. Most of its attenuation is provided by the iodine component due to its higher atomic number and physical density. Also the K-edge of iodine occurs at 33.2 keV, at the center of the energy spectrum used for most diagnostic studies. Similarly, the barium contrast agent used in abdominal studies contains 450 grams of barium sulfate (Ba $SO_4$) in 25 milliliters of water to give a suspension with a physical density of 1.20 gm/cm$^3$. The K-edge of barium occurs at 37.4 keV, again lying near the center of the energy spectrum used for abdominal studies.

One can maximize the contrast of iodine and barium contrast media by imaging with x-ray photons just slightly above the k-edge of the contrast agent. This is achieved by "shaping" the x-ray spectrum by an additional metallic filter with a k-edge higher than the k-edge of the contrast agent. The metallic filter attenuates photons at energies higher than its k-edge but transmits a larger proportion of photons with energies just below its k-edge (Figure 4-8). If the k-edge of the filter is higher than the k-edge of the contrast agent, a large proportion of the transmitted photons will fall over the k-edge of the contrast agent. Since these photon energies are chosen to fall in a region of maximum attenuation for the contrast agent, they will maximize its contrast. As you might deduce, metals with slightly higher atomic numbers than the contrast agent are useful as x-ray beam filters in these applications since they also have slightly higher k-edges. Therefore rare earth metals such as samarium (Sm) and cerium (Ce) are commonly used to filter the x-ray beam in contrast studies involving iodine or barium. Because of the principle of their operation, they also are called "k-edge" filters. In Figure 4-8 note that the 120-kVp x-ray beam contains many high energy x-rays where iodine has a low attenuation coefficient. A 60-kVp spectrum contains many lower energy photons, below the iodine k-edge. A rare-earth filter (e.g., samarium; k-edge = 46.8 keV) removes lower and higher energy photons and shapes the 60-kVp spectrum for improved absorption by iodine.

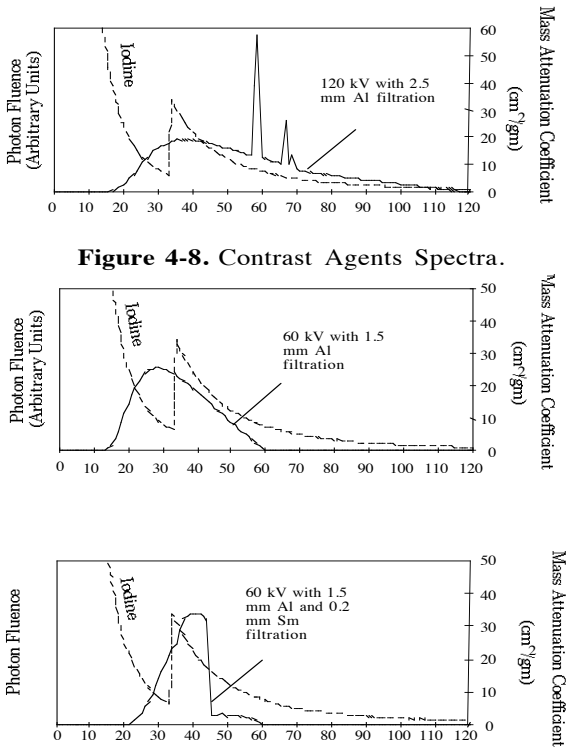



**Figure 4-8.** Contrast Agents Spectra.

A third contrast agent, one that reduces rather than increases x-ray attenuation, is air. Prior to the advent of computed tomography, radiographic images of brain structures were obtained after injecting air into the cerebral ventricles through a catheter inserted in the spinal column. The introduction of air displaced cerebral spinal fluid in the ventricles, which otherwise has essentially the same x-ray attenuation properties as the surrounding brain tissue. Without the introduction of a contrast agent, the ventricles could not be visualized with standard x-ray techniques, nor could the various white and gray matter structures in the brain be differentiated using plane-film radiography. The air injected into the ventricles displaced the cerebral spinal fluid, allowing the radiologist to visualize the shape of the ventricles. Any distortion in the shape suggested the presence of a tumor or other abnormality. In the past 15 years, computed tomography and MRI have all but completely eliminated the use of this technique, pneuomoencephalography. Computed tomography and MRI have been phenomenal advances for studies of the brain, particularly since pneuomoencephalography was quite painful and prone to complications, in rare instances including death. However, it remains an example of how air can be used as a contrast agent to visualize structures that normally would be invisible in a radiograph.

Another study where air is used as a contrast is imaging in the gastrointestinal (GI) tract. In the so-called "double contrast study", a barium contrast agent is introduced. This is followed by injecting air into the GI tract to displace the bulk of the barium contrast agent, leaving behind a thin coating of barium on the inner surface lining of the GI tract. This allows the radiologist to observe the intricate structure of the GI lining to look for ulcerations.

Of course, the property of air that makes it a useful contrast agent is its low physical density. Since both the photoelectric and Compton linear attenuation coefficients of a material are directly proportional to its physical density, air maintains a low linear attenuation coefficient with respect to any other material found in the body (Fig 4-9).

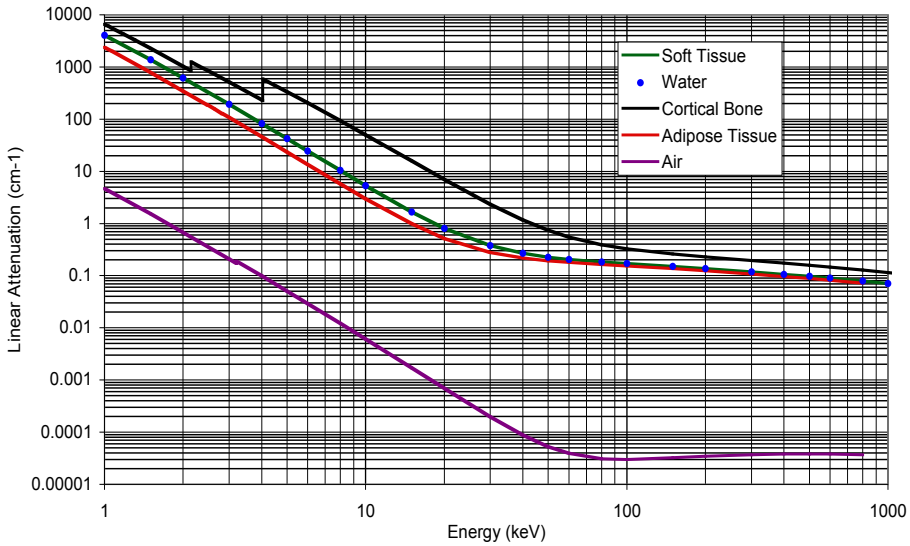



**Figure 4-9**. Linear Attenuation Coefficients ($cm^{-1}$).

## 4.5 Detector Contrast

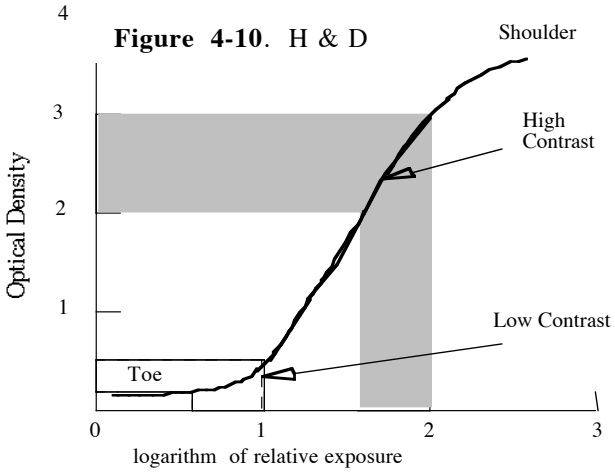
In the previous section, we presented methods by which radiographic or subject contrast can be maximized. However, a human observer never directly responds to the radiographic contrast since we are not equipped with biological sensors that can detect photons with energies in the diagnostic range. Instead the x-ray beam is translated to an intermediate usable form by a film-screen cassette or an electronic detector. Of course, the choice of the detector has an important bearing on the final image contrast, which we observe, making these properties important for us to consider. We first will discuss the properties of a film-screen cassette that contribute to contrast, then discuss electronic detectors.

Contrast With Screen-Film System

An x-ray beam transmitted through a patient enters a film-screen cassette and strikes the intensifying screen, which gives off light to expose the photographic emulsion. The amount of light from the intensifying screen depends on the amount and energy of the photons as well as the chemical composition of the intensifying screen and its thickness. The chemical composition of the intensifying screen determines first how much radiation is absorbed by that screen, and secondly how much light is given off for each unit of radiation that is absorbed.

The choice of the intensifying screen can affect detector contrast. Since different regions of the body contain tissues with different elemental compositions, the energy spectrum of the beam emerging from the body and striking the image receptor can be different from one region of the body to another. In a few cases, this can contribute to differences in contrast. For example, if one region of the body contains an iodinated contrast agent then x-ray photons with energy below the iodine k-edge will present a small radiographic (or subject) contrast, while those just above the iodine k-edge will present a much larger radiographic (or subject) contrast. A screen that is more sensitive to the photons above the k-edge therefore will generate an image with a higher contrast than a screen that is more sensitive to photons

below the k-edge. In most cases, this is not an important effect for film-screen radiography. Therefore one typically chooses the screen to be most sensitive to the entire spectrum of the beam emerging from the patient to minimize image noise rather than choosing the screen to be sensitive to one particular part of the energy spectrum to maximize contrast.




Second, beyond the film's linear range, additional exposure of the film results in a smaller increase in its optical density. The photographic film therefore has both a toe (at low exposure levels) and shoulder (at high exposure levels) where the H&D curve has a smaller gradient than that found in the mid or linear portion. Both the toe and the shoulder of the H&D curve are regions of low film contrast, so film exposure must be chosen carefully to fall in the film's linear region though this is not always possible across the entire field of view of the image (Figure 4-11)

In comparison to the intensifying screen in a film-screen system, the film can have a dramatic effect on image contrast. A screen can respond linearly by emitting light proportional to the x-ray exposure over multiple decades of exposure. However, photographic film does not respond similarly to the light emitted by the screen. Rather, its characteristic (or "H and D") curve relating film exposure to film density is a sigmoid ("S-shaped"). This means that, first, a certain level of light exposure has to be reached before the density of the film increases linearly with its exposure (Fig 4-10).

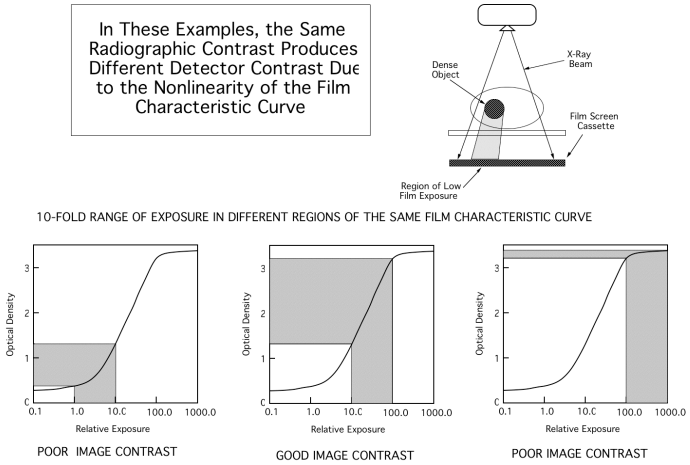
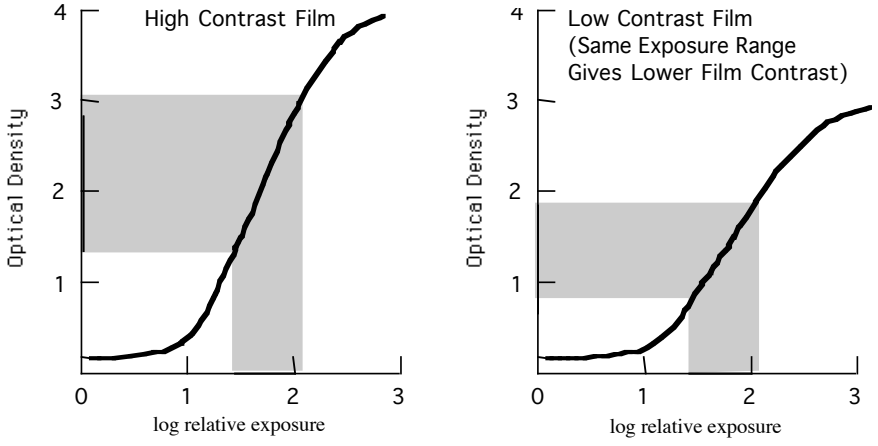



**Figure 4-11**. Film Contrast

**Figure 4-12**.

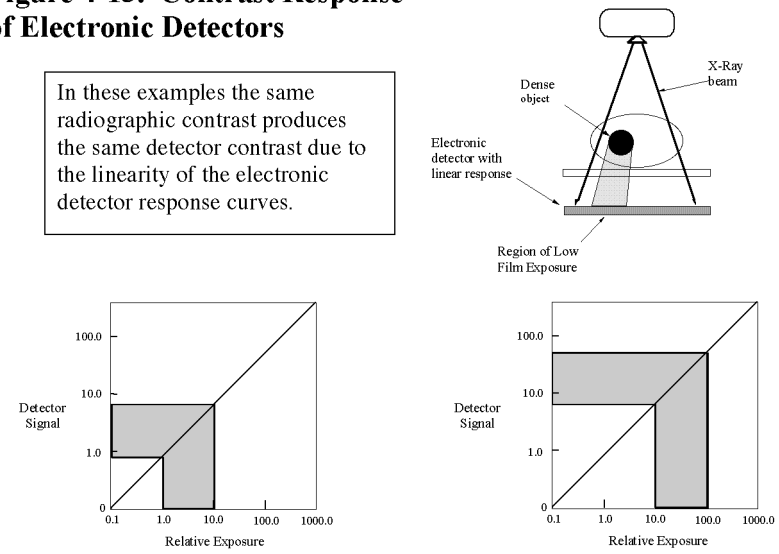



Another important determinant of detector contrast with a film-screen system is the film-gamma or film-gradient, which is the slope of the film characteristic curve in its linear region. If presented with the same radiographic contrast, a film with a larger gamma will produce a larger detector contrast than one with a smaller gamma (Fig 4.12). A principal disadvantage of film with a large gamma is diminished latitude. This means a smaller range of exposures producing a linear response than a film with a small gamma. Therefore studies such as chest radiography, which produce wide variations in photon fluence between the lung and the mediastinum or abdomen (high radiographic

image latitude), often utilize films with higher latitude, i.e. with lower contrast. Alternatively, studies such as mammography use high-contrast films since the images are obtained with compression of the breast to help minimize unwanted exposure latitude. **When a film-screen system is used, the characteristic curve of the film is the most important determinant of detector contrast**.

4.5.2 Contrast with Electronic Detectors

**Figure 4-13. Contrast Response of Electronic Detectors**



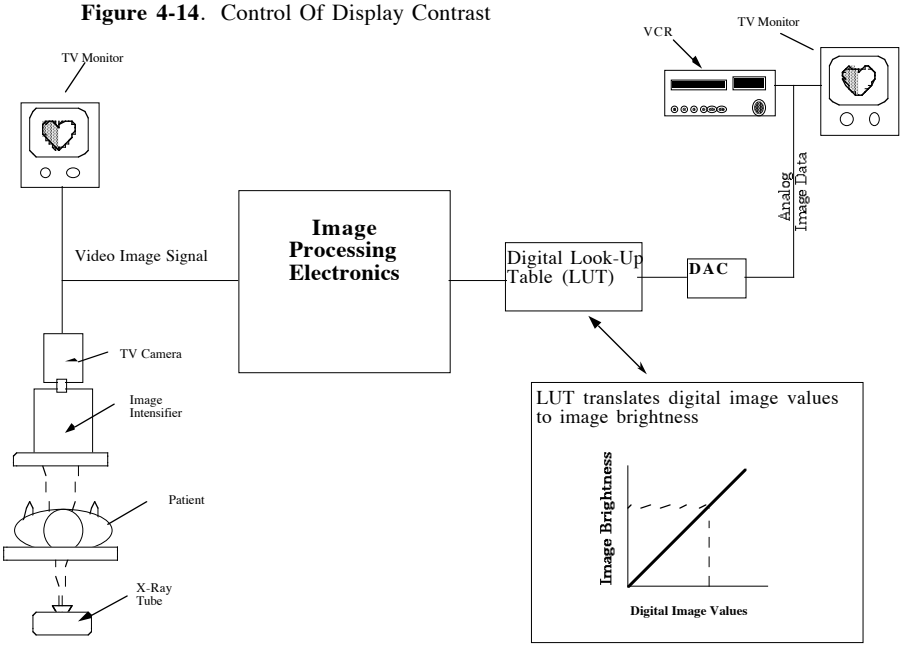

There are several different materials used in electronic detectors. Common materials include sodium iodide, bismuth germinate, calcium tungstate, cadmium tungstate, calcium fluoride, cesium iodide, high purity germanium, xenon gas, and arrays of high-Z solid state detectors. As in the case of intensifying screens, each of these materials has an individualized spectral response. That is, depending on its composition and thickness, each material will absorb x-ray photons as a function of energy to various degrees. In the case of a scintillation detector, the light output φ from the scintillator depends on the total amount of energy absorbed by the detector material. We can calculate this value using the equation

$$\Phi = K \int S(E) \left\{ 1 - \exp\left[ -\left(\frac{\mu}{\rho}\right) \rho x \right] \right\} E dE \qquad (4\text{-}3)$$

where S(E) represents the energy spectrum of the x-ray beam in terms of the number of photons for each unit of energy in the energy range of the beam, and K represents the fraction of absorbed x-ray energy that is converted to light. Whatever the active detector material, electronic detectors differ from photographic detectors, because unlike film their response is linear across a greater range of radiation exposure. While photographic detectors generally have a linear response to radiation exposure that spans one to two orders of magnitude, an electronic detector will often have a linear response spanning five orders (Figure 4-13). As discussed in the previous section, if an object presents the x-ray beam with a wide range of attenuation, the resulting exposures may exist outside of the linear region of the film characteristic curve (Figure 4-10). This reduces contrast and is an inherent limitation of film. In comparison, the linearity of the electronic detector response maintains the contrast in the radiographic signal, when imaging an object that has a large range in x-ray attenuation. This property is often touted as an important advantage when electronic detectors instead of film are used for imaging objects having a wide range of attenuation.

Display Contrast

**Figure 4-14**. Control Of Display Contrast




The availability of electronic images allows us to use electronic enhancement to increase image contrast in a way that is not available with film-screen systems. Electronic images are stored as arrays of digital numbers (we will call them pixel values for simplicity) rather than in terms of optical density as in a conventional radiograph. In all electronic image processing systems, the translation from digital image value to image brightness is performed through a "look-up table" (LUT) which specifies the image brightness that corresponds to a particular digital image value (Figure 4-14).

The operator controls the contents of the "LUT" by adjusting the "window" and "level" of the image display (Figure 4-15). The level determines which pixel value maps to the mid grey level. The window determines the range about this pixel value to map from black to white for display. For example, the operator can specify that a narrow range of pixel values (i.e. a "narrow window") for full display range from black to white giving a high contrast image display. Alternatively, the operator can specify that a wide range of pixel values (i.e. a "wide window") for the full display range from black to white, giving a low contrast image. A narrow window also can be moved up and down through the range of image values (i.e. adjusting image "level") so that only structures of low- or high-pixel values are selectively displayed at high contrast. All of this can be performed under operator control.

Therefore, the availability of electronic imaging systems provides flexibility in image contrast that is not available with conventional film-screen systems. The availability of electronic images allows us to display images with color. In most planar imaging applications, color is not used because the transitions from one hue to the next can create false borders visually that distract the observer. However, color displays can be very helpful in parametric or functional imaging where pixel values represent a definite quantity such as blood flow, transit time, or some other physiological measurement derived from images. Color simplifies comparing pixel values in distant parts of an image where visual comparison often fails. Color is also used in tomographic image studies, especially in radionuclide imaging (SPECT and PET) to help visualize edges. Also, color rendering is used in some surface models derived from CT and MR images in an attempt to give a more realistic appearance or to show deep structures hidden behind superficial ones. However, it is important to note that these electronically generated color images should not be confused with the normal conception of real world colors, rather they are “pseudo color” images.

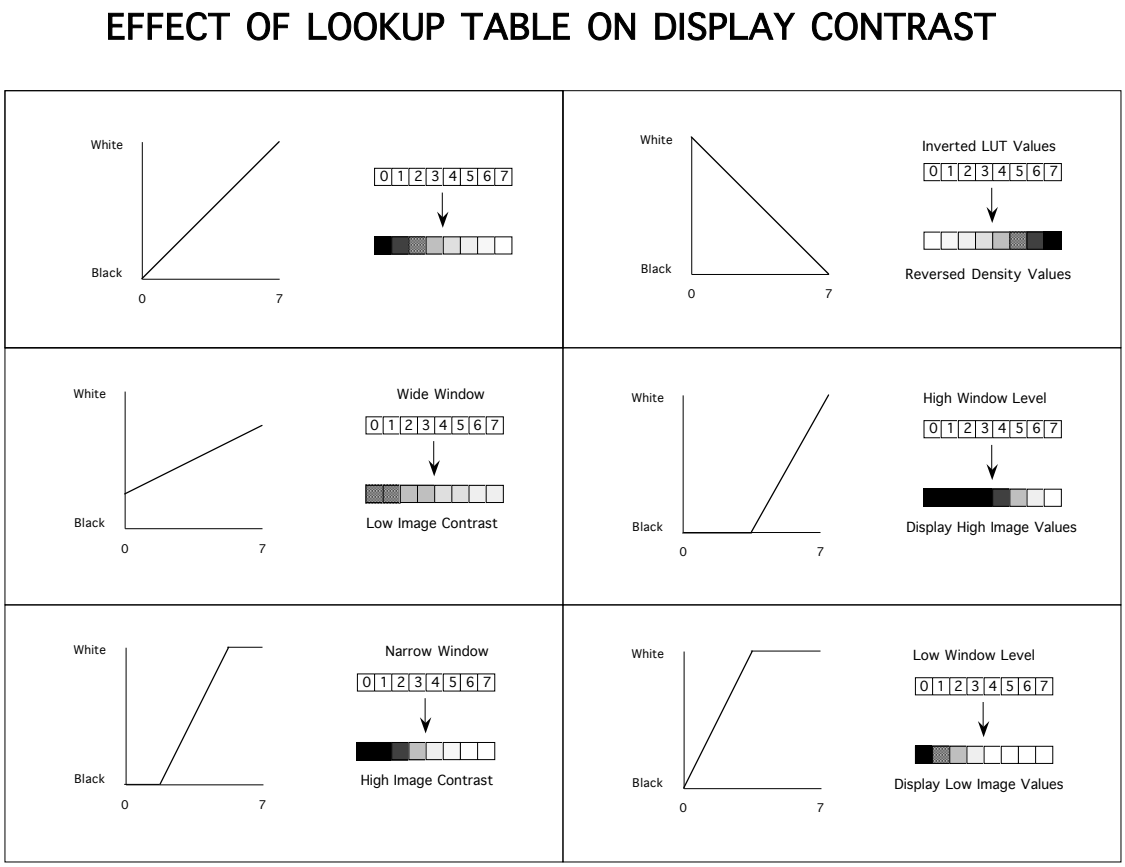



**Figure 4-15**. The digital look-up table gives the relationship between an input digital number (horizontal axis) and an image brightness (vertical axis). By changing the characteristics of the look-up table, the display contrast can be altered drastically.

## 4.6 Scattered Radiation and Image Intensifier Veiling Glare

Finally, there are three related physical perturbations to radiologic images that act to decrease contrast, scattered radiation, fog in the film, and veiling glare in the image intensifier. We will discuss scatter radiation first and briefly discuss the similar effects of veiling glare. Fog in film can be minimized by proper storage and handling, but otherwise cannot be avoided and will not be discussed further in this chapter.

### Scattered Radiation

Scattered radiation is present in all radiographic studies, and is an important consideration especially in the majority of diagnostic examinations that use broad area beams, instead of narrow (fan or pencil) beam geometry. Measurements by Sorensen show that, in the standard chest radiograph, scatter radiation can account for 50% of the radiographic signal behind the lungs and up to 90% of the signal behind the mediastinum and diaphragm when no scatter rejection techniques are used.

The image formation process in diagnostic radiology essentially captures a radiographic "shadow" created by the body of x-rays from a point source. The accuracy of this "shadow" depends on the photons being highly directional. However, scattered radiation is not emitted by a single point source.

Rather, it strikes the film-screen cassette from random directions and carries little useful information, unlike the directional primary photons that arise from the source. A useful way to describe the amount of scatter in a radiographic signal is the scatter fraction F defined as

$$F = \frac{\text{exposure contributed by scattered radiation}}{\text{exposure contributed by primary and scattered radiation}} \quad (4\text{-}4)$$

Sorenson has shown that the contrast reduction due to scatter radiation is related to the scatter fraction F according to the equation

$$C_{sc} = C(1\text{-}F) \quad (4\text{-}5)$$

where

$C_{sc}$ = contrast in the presence of scatter
C = contrast with no scatter in the image
F = scatter fraction

When the scatter fraction is 90% as it can be in chest radiology, the reduction in contrast due to scatter is also 90% and one obtains only 10% of the contrast that is available in the image without scatter (Figure 4-16). It is therefore important to reduce the amount of scatter for any imaging procedure.

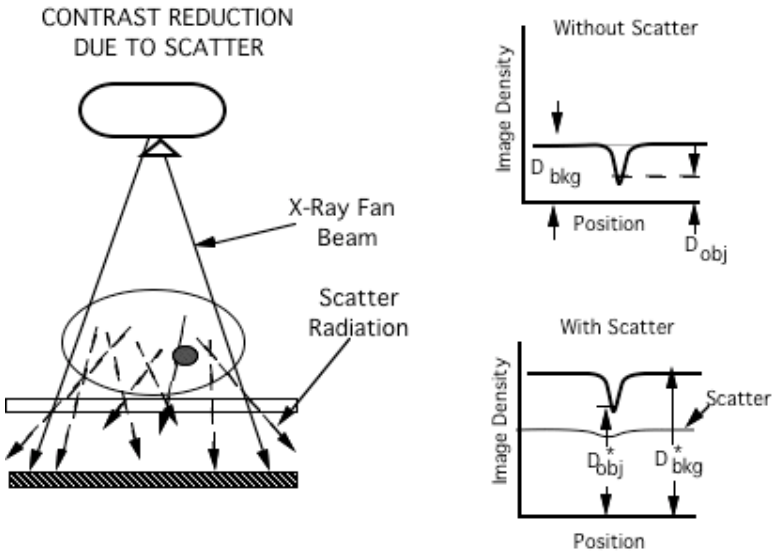
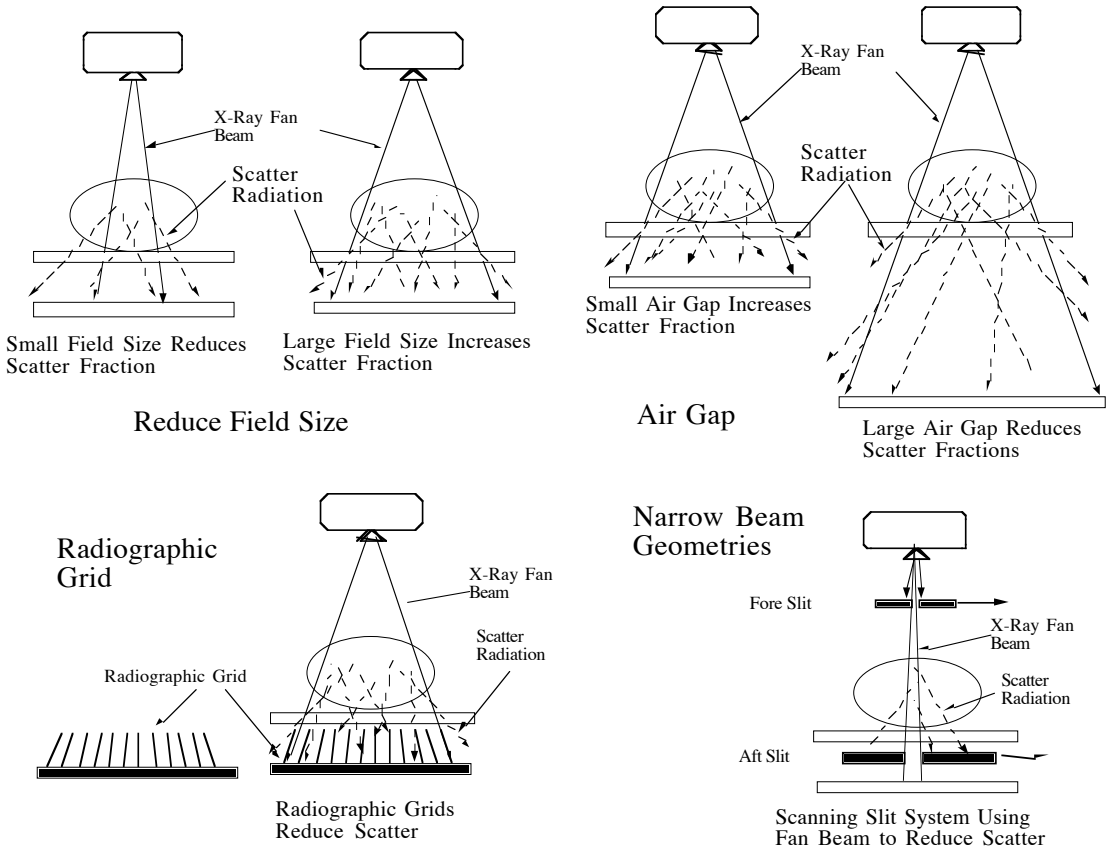



**Figure 4-16**. Scatter and Contrast

There are a number of ways to reduce scattered radiation to improve radiographic contrast (Figure 4-17). We will discuss these techniques in detail in later sections and only review them conceptually here. Scatter can be reduced by geometrical means, by using a small field size or by having an "air gap" between the patient and the image receptor. In these cases, because the scatter does not travel in the direction of the primary x-ray beam, it has a lower probability of striking the image receptor with a small field size or large distance between the patient and detector. In addition, when a small field size is irradiated, less tissue volume is exposed and therefore less scatter radiation is generated. It is always good radiological practice to limit the radiation exposure to as small an area (field of view) as possible consistent with the clinical requirements of the study. This reduces the integrated radiation dose delivered to the patient and helps recover contrast lost due to scattered x-rays.

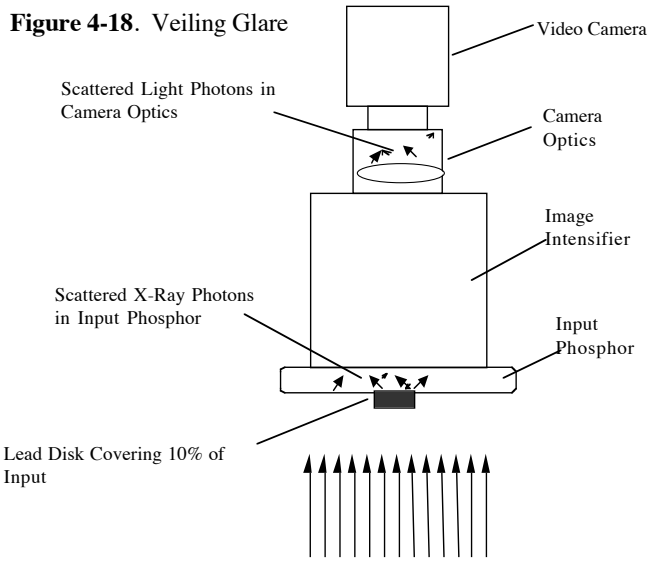
## **Figure 4-17**. Scatter Reduction Techniques




There also are external means by which scatter can be removed from the image. The most common technique uses a device called a grid that is analogous in concept to venetian blinds. A grid contains thin parallel strips of lead embedded in aluminum or plastic with the strips aligned with the direction of the x-ray beam emerging from the x-ray tube. The grid is placed between the patient and the image receptor. Since its plates are parallel to the direction of the primary x-ray beam, x-ray intensity of photons traveling in a direction different than the primary beam is reduced by the lead strips, thereby reducing scattered radiation.

### Image Intensifier Veiling Glare

**Figure 4-18**. Veiling Glare




Veiling glare (Figure 4-18) arises from the scattered x-rays at the input and output windows of the image intensifier as well as scatter of visible light in the input and output optics of the image intensifier and television camera.

Indeed, veiling glare occurs to a certain extent in all electro-optical imaging systems used in

diagnostic radiology. Its effect is similar to that of scattered radiation in that veiling glare reduces contrast. Its contrast-reducing effect can be quantified in terms of the contrast ratio, which is defined as the image intensifier output at the center of the image intensifier divided by its output when 10% of its input area is blocked by a lead stop (so that the output is due entirely to veiling glare). Ideally, one wants no veiling glare, corresponding to an infinite contrast ratio. Conventional image intensifiers using cesium iodide (CsI) phosphors have a contrast ratio of 17:1 while those with a fiber-optic output window and very thin titanium input window have contrast ratios of 35:1. If an image intensifier has a contrast ratio of R, an ideal contrast $C_o$ will be reduced to a contrast $C_{vg}$ with veiling glare where

$$C_{vg} = C_o \left[\frac{R}{R+1}\right] \qquad (4\text{-}6)$$

For example, if an image intensifier has a contrast ratio of 20:1, then an ideal contrast of 10% will be reduced to 9.5% (about a 5% loss) due to the effects of veiling glare.

## 4-7. Physical Determinants of Contrast for Several Other Imaging Modalities

**Nuclear Medicine -** Imaging radiotracers with a gamma camera (functional imaging).

Subject Factors:

Organ/tissue uptake pattern
- target-to-background ratios for radiolabelled tracers
- normal vs. abnormal tracer uptake

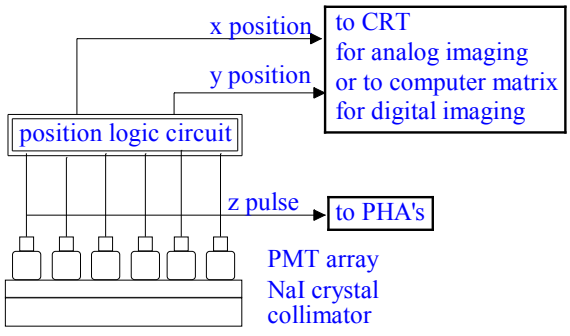
Emission escape probability
- gamma energy - high vs. low
- location - shallow vs. deep
- body size - small vs. large

Detector Factors:

Crystal
- sensitivity (thickness, energy, energy resolution)
- spatial resolution (thinness, PMTs, electronics)

Spectrometer
- window/level
- scatter rejection

Collimator
- resolution vs. distance
- sensitivity vs. resolution
- scatter



**Magnetic Resonance Imaging** - Conventional MRI

Subject Factors:

The most important subject factors are the T1 and T2 relaxation times and proton density. Clinical MRI images are of signals from hydrogen protons in tissue water. The magnetic moment of these protons makes them sensitive to RF signals of ~43 MHz at one Tesla.

T1 - varies with field strength ($B_0$) (long for CSF, short for fat, intermediate for other tissues)

T2 - minimally affected by $B_0$ (~$1/10^{th}$ the T1) (ordering it tissues similar to T1)

PD - lowest for air, low for bone, high for water, close to that of water (but lower) for most tissues

General

T1 difference important for GM/WM contrast in brain images
T2 difference important for pathology (usually edema)
PD determines potential signal strength

Detector Factors:

Signals from MR images are intrinsic, coming from the protons of water. The imaging system consists of an RF transmitter with proper coils, gradient coils for spatial encoding, and an RF receiver with proper coils. The RF signals from the receiver are sorted into "k" space, which is a Fourier transform of signals from the object (subject or patient). An inverse Fourier transform is taken and the magnitude calculated to form MR images.

"Spin-Echo" pulse sequences are commonly used for clinical studies.

MR image contrast is determined mainly by the repetition time (TR) and the echo time (TE), which are console selectable parameters for spin-echo pulse sequences.

In general image contrast is weighted by either T1 (T1W) or T2 (T2W) relaxation times, or proton density (PDW). The following table summarizes the relationship between the acquisition parameters and the weighting mechanism used to develop contrast in MRI.

| Parameter | T1W | T2W | PDW |
|---|---|---|---|
| TR (msec) | SHORT (450-850) | LONG (2000+) | LONG (2000+) |
| TE (msec) | SHORT (10-30) | LONG (>60) | SHORT (10-30) |

In T1W images tissues with shorter T1 times (e.g. fat) have higher signals.
In T2W images tissues with longer T2 times (e.g. CSF) have higher signals.
In PDW images tissues with greater proton densities have higher signals.

Reconstruction Algorithms: Number of excitations, number of phase encodes, array size

## Computed Tomography -

Subject Factors:

CT Number $= [(\mu - \mu_w)/\mu_w] \bullet 1000$ (contrast in $\mu$ relative to water)

$\mu = \mu_m \rho$ (mass attenuation coefficient x mass density)
Contrast $= [(\mu_m \rho)_1 - (\mu_m \rho)_2)] / (\mu_m \rho)_2$
If $\mu_{m1} \approx \mu_{m2}$ then Contrast $\approx (\rho_1 - \rho_2)/\rho_2$
If $\rho_1 \approx \rho_2$ then Contrast $\approx (\mu_{m1} - \mu_{m2})/\mu_{m2}$

Energy
KVp - 125 with heavy beam filtering

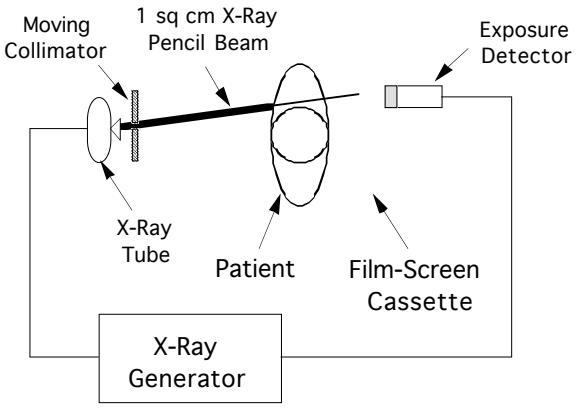
Detector Factors:

Collimators
between plane resolution - slice thickness
in plane resolution
pre and post collimation
scatter

Number and Spacing
Resolving time
Reconstruction Algorithms: Reconstruction filters, array size

## CHAPTER 4: HOMEWORK PROBLEMS

1. Film Compensation System: Several laboratories are investigating "exposure compensation" systems that modulate the exposure to the patient so that the film exposure is relatively uniform.




In such a system (diagrammed above), the x-ray tube output is collimated to a pencil beam having a 1 $cm^2$ cross-sectional area. The x-ray beam is swept from left to right across the patient to expose a film-screen cassette. At each location, a detector placed behind the film-screen cassette monitors the exposure so that when it reaches a certain constant level, a feedback circuit terminates the exposure. The x-ray beam then is moved to an adjacent point on the film and the exposure cycle is repeated.

(a) To what level should the film be exposed to maximize image contrast?

(b) What are the effects on contrast of this technique in comparison to conventional imaging methods that deliver a uniform exposure to the patient rather than to the image receptor?

(c) What are the effects on noise (quantum statistics) as well as on the scatter distribution of this method in comparison to conventional imaging techniques?

(d) Assume that the optimal film exposure level in <u>air</u> is achieved in time $\tau$. If the patient thickness is equivalent to x cm of water, show that when the beam passes through the patient, the optimal film exposure level is achieved in time t where

$$t \sim \tau \exp(\mu_{water} x)$$

(e) Briefly outline the design of the electronic feedback system to implement exposure compensation in a clinically usable system.

2. Assume that a 1-mm diameter artery contains a 10 mg/cm$^3$ concentration iodine solution.

   (a) Calculate the approximate radiographic (subject) contrast assuming an effective attenuation coefficient of 15 cm$^2$/gm if no scatter is present. Make any reasonable assumptions you need to answer this question.

   (b) What is the radiographic contrast if the scatter fraction is 0.5?

3. In a quantum limited imaging system, a low contrast object is just distinguishable from its surroundings.

   (a) If the object has an area of a 1-cm$^2$ and the measured exposure is $10^{-5}$ R, use the Rose model to estimate the radiographic contrast. Make any reasonable assumptions you need to answer this question.

   (b) If the object has an attenuation coefficient of 10 cm$^2$/gm, estimate the product of density and thickness in the object which is just distinguishable from its surroundings.

4. Consider two imaging situations using contrast agents with electronic detectors. In one case, a detector of cesium is used to image an iodine contrast agent. In the other case, a detector of iodine is used to image a cesium contrast agent.

   (a) In which case would you expect the better image contrast? Why?

   (b) In which case would you expect the better signal-to-noise ratio at a given patient exposure level? Why?

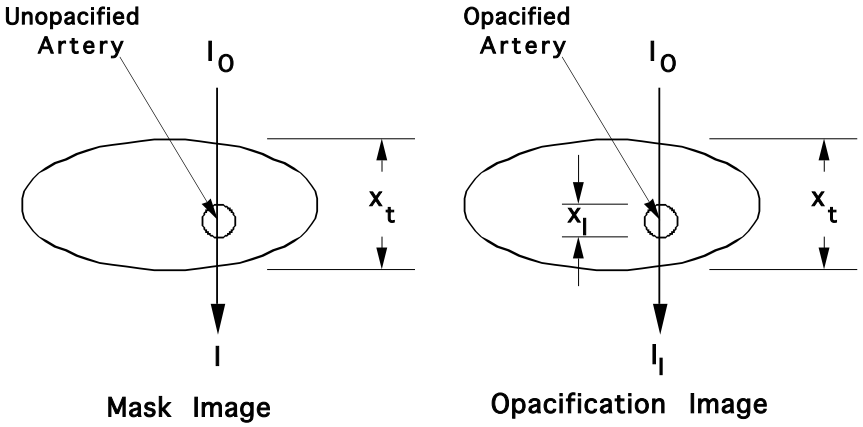
   (c) Which of these two hypothetical situations is more realistic in terms of commonly available contrast agents and detector materials?

5. Polycythemia is a disease characterized by an abnormal increase in the number of circulating red blood cells. In its normal function of breaking down hemoglobin from dead red blood cells, the livers of patients with polycythemia accumulate abnormal amounts of iron.

   (a) Ignoring patient radiation dose considerations and knowing that the k-edge energy of iron is 7.11 keV, describe how you would choose the kVp and the filter for the x-ray tube to maximize liver contrast for patients with polycythemia.

   (b) Briefly comment whether your answer in (b) is clinically realistic when patient radiation dose is considered. A short calculation may help to clarify your answer.

6. Digital subtraction angiography (DSA) is a technique using a digital fluorographic system. In DSA, one image (the "mask image") is obtained before an iodinated contrast agent is injected into the circulatory system. A second image (the "opacification image") then is obtained after injection of the contrast agent. The images are logarithmically transformed, then subtracted to remove anatomical background structures to isolate the contrast agent in the circulatory system (typically arteries and cardiac chambers). The geometry of the two images is summarized in the diagrams below.




where

$I_0$ = incident photon fluence for both the mask and opacification images
I = photon fluence detected in the mask image
$I_I$= photon fluence detected in the opacification image
$\mu_I$ = linear attenuation coefficient of iodine within the artery
$x_I$ = equivalent thickness of iodine from the contrast agent in the opacification image
$\mu_t$ = linear attenuation coefficient of tissue
$x_t$ = equivalent thickness of tissue in both the mask and opacification images

(a) Show that for the mask image

$$I = I_0 \exp(-\mu_t x_t)$$

and that for the opacification image

$$I_I = I_0 \exp[-\mu_t x_t - \mu_I x_I).$$

(b) If we define the logarithmic difference image L to be

$$L = \ln(I_I) - \ln(I)$$

show that

$$L = -\mu_I x_I.$$

Briefly discuss assumptions and how this result shows that logarithmic subtraction removes anatomical background structures from the DSA image.

(c) Define the scatter-to-primary ratio f as

$$f = \frac{\text{exposure contributed by scattered radiation}}{\text{exposure contributed by primary radiation}}$$

In the presence of scatter radiation, if f is the scatter-to-primary ratio, the photon fluence in the mask image is contributed by both the primary (unscattered) and the scattered components. In this case, show that the detected photon fluence is

$$I = I_0 \exp(-\mu_t x_t) + f\, I_0 \exp(-\mu_t x_t)$$

Assuming that the scatter field does not change significantly by the addition of a small amount of iodine, show that for the opacification image

$$I_I = I_0 \exp(-\mu_t x_t - \mu_I x_I) + f\, I_0 \exp(-\mu_t x_t) .$$

(Hint: Part (c) follows directly from the definition of the scatter-to-primary ratio. The solution is very simple. Do not make it too difficult for yourself.)

(d) For a scatter-to-primary ratio f, show that in the presence of scatter, the logarithmic difference image is

$$L = \ln\left(\frac{e^{-u_I x_I} + f}{1+f}\right)$$

which reduces to

$$L \cong -\frac{\mu_I x_I}{1+f}$$

for small values of $\mu_I x_I$ . Discuss the importance of this result in terms of the image contrast that is obtained in digital subtraction angiography.

7. Dual-energy imaging is a technique in which two images are obtained with different energy spectra with the purpose of separating two materials having dissimilar atomic numbers that are both present in the imaging field. In most realistic systems, the spectra used for image acquisition are polyenergetic bremsstrahlung spectra, but for purposes of this problem, we can approximate the beams as being monoenergetic with energies $E_1$and $E_2$. Assume that

$I_{01}$= incident photon fluence at energy $E_1$
$I_1$= detected photon fluence at energy $E_1$
$I_{02}$= incident photon fluence at energy $E_2$
$I_2$= detected photon fluence at energy $E_2$

(a) If the imaging field contains only bone (b) and soft-tissue (t), and if

$\mu_{b1}$= linear attenuation coefficient of bone at energy 1
$\mu_{b2}$= linear attenuation coefficient of bone at energy 2
$\mu_{t1}$= linear attenuation coefficient of soft tissue at energy 1
$\mu_{t2}$= linear attenuation coefficient of soft tissue at energy 2

show that the thickness of bone $x_b$ and the thickness of soft tissue $x_t$ are equal to

$$x_b = \frac{\mu_{t1}\ln\left(\frac{I_{O2}}{I_2}\right) - \mu_{t2}\ln\left(\frac{I_{O1}}{I_1}\right)}{\mu_{t1}\mu_{b2} - \mu_{t2}\mu_{b1}} \qquad x_t = \frac{\mu_{b1}\ln\left(\frac{I_{O2}}{I_2}\right) - \mu_{b2}\ln\left(\frac{I_{O1}}{I_1}\right)}{\mu_{b1}\mu_{t2} - \mu_{b2}\mu_{t1}}$$

(b) Briefly discuss how the results in (a) allow you to separate bone and soft tissue when imaging the human body using dual-energy techniques.