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X-ray Image Formation

Lecture 5

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1- X Ray Image formation

1-1 Components of an imaging system

The principal components of a system for X ray projection radiography are illustrated in fig. 6.1. Of these components, the grid and the automatic exposure control (AEC) are optional, depending on the imaging task. Further components such as shaped filtration, compression devices or restraining devices may be added for special cases. the X ray tube and collimation device are described in chapter 3 . When considering such systems, the concept of an ideal imaging task is often useful, as illustrated in fig. 1. When considering the ideal imaging task — the detection of a detail against a uniform background — the ideal X ray spectrum is monochromatic when the three constraints of patient dose, image quality and X ray tube loading are considered. Any particular projection may consist of more than one such task, each with a different ideal monochromatic energy. the choice of X ray spectrum for each task is, therefore, always a compromise, so that the actual bremsstrahlung and characteristic radiation spectrum provide the best approximation to the ideal monochromatic spectrum for the particular projection.

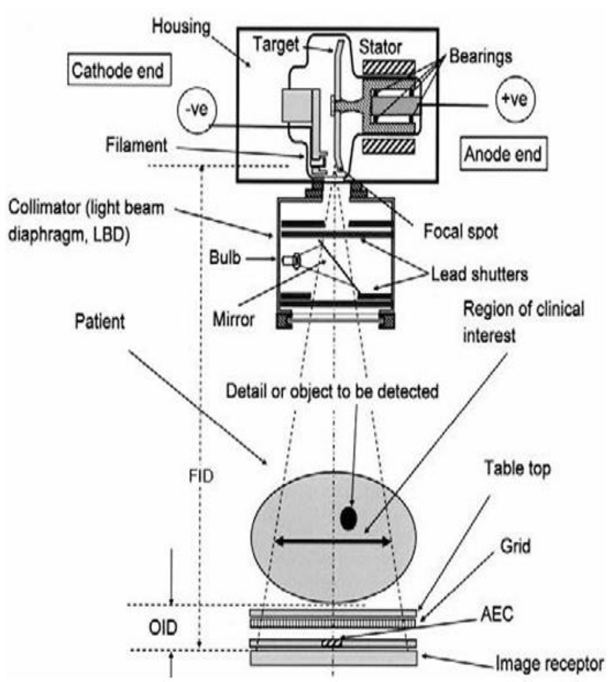

FIG.1: Components of a projection radiography system, including an ideal imaging task, for the detection of a detail against a background. FID: focus to image distance; OID: object to image distance.

considering an ideal imaging task, as illustrated in fig. 1, contrast may be defined simply as $C = \Delta B/B$, where B is the image brightness (or shade of grey) in a background region and ΔB is the difference in brightness for a small detail. for small values of ΔB, linearity of brightness with X ray intensity *(I)* is assumed, so the contrast is $\Delta I/I$. this is generally valid for a particular monochromatic spectrum. for a real polychromatic spectrum, a monochromatic spectrum with the average energy of the actual spectrum may be used as an approximation, or the result can be integrated over all spectral energies. since the X ray intensity is related to thickness by the attenuation law, it follows that the primary contrast for a detail of thickness x_d and linear attenuation coefficient μ_d embedded in a material of linear attenuation coefficient μb is given by:

$$C_{\mathrm{P}} = 1 - e^{-(\mu_{\mathrm{d}} - \mu_{\mathrm{b}})x_{\mathrm{d}}} \qquad \ldots\ldots\ldots\ldots\ldots\ldots\ldots \quad (1)$$

to find the average contrast for a particular detail, eq. (1) should be integrated over the detail. for thin spherical details (e.g. microcalcifications in mammography, solitary pulmonary nodules in chest X rays), this is straightforward and results in a contrast that is 2/3 the contrast obtained for a ray passing through the centre of the detail. With regard to the relationship between the linear attenuation coefficient and the mass attenuation coefficient , contrast will exist for details that differ in mass attenuation coefficient, or in density, or both. the contrast will depend on the thickness of the detail, but not the thickness of surrounding tissue. since the values of μ reduce as photon energy increases, the contrast is seen to be inversely related to the kV setting. thus, kV may be considered to be the contrast control, where contrast is strictly the detail contrast. for screen film imaging, the difference in optical density (OD) due to the detail is proportional to the subject contrast multiplied by the gamma of the screen film system for digital image receptors, the relationship is more complex, since the contrast of the displayed image is independently adjustable. if the energy absorbed in a small region of the image receptor due to primary rays is E_p , and that due to secondary rays is E_s , then scatter may be quantified by the scatter fraction:

$$SF = E_s/(E_p + E_s) \qquad \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots(2)$$

or by the scatter to primary ratio:

$$SPR = E_s / E_p \qquad \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots(3)$$

the relationship between the two is:

$$SF = ((SPR–1) + 1)–1 \qquad \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots(4)$$

in the presence of scattered radiation, eq. (1) becomes:

$$C_p = 1 - e^{-(\mu_d - \mu_b)x_d} \frac{1}{1+SPR} \qquad \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots(5)$$

clearly, minimization of scatter is important, leading to the use of antiscatter techniques. accurate collimation to the region of clinical interest also minimizes scatter, as well as reducing the dose to the patient.

1-2 Geometry of projection radiography

from fig. 6.1, it is clear that the primary effect of projection radiography is to record an image of a 3-D object (the patient) in 2-D, resulting in superposition of the anatomy along each ray. this leads to a number of effects that need to be considered in the design of equipment, the production of the images and their interpretation. in particular, for each projection there will be a region of clinical interest, somewhere between the entrance and exit surface of the region to be imaged. considerable training and experience is required for the radiographer to choose correctly the geometrical variables to image this region, based on superficial visible or palpable landmarks. these variables include the fiD, oiD, projection direction (lateral, craniocaudal, etc.) or angulation, centring point and beam collimation area. in some cases, the correct projection of joint spaces also needs to be considered.

1-3 Effects of projection geometry

1-3-1 Superposition as noted in section

radiographs are a 2-D representation of a 3-D object. this superposition leads to a significant loss of image contrast, which provided one of the prime motivations for the development of CT scanners. superposition also leads to loss of all depth information, and ambiguity in the relative sizes of objects at different depths. furthermore, it directly overlays objects in such a way that it can become difficult or impossible to distinguish one from the other, or even to identify some of the objects

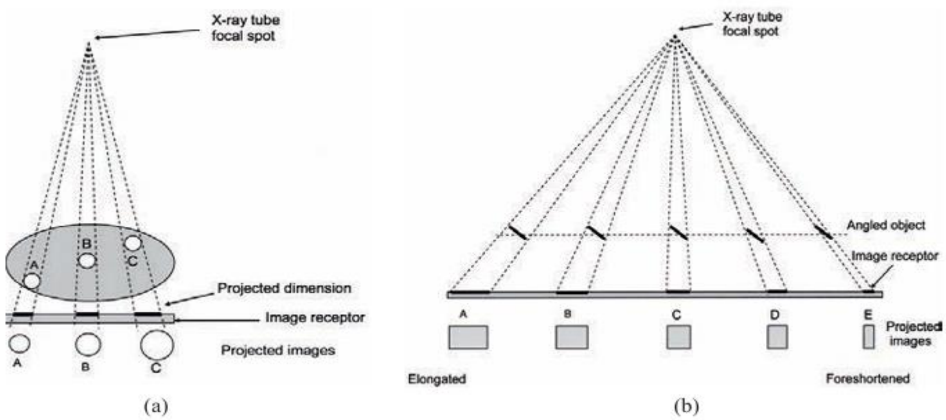

FIG. 2: (a) Effect of depth of objects on their projected size; (b) effect of angulation on the projected length of an angled object.

1-2-2 Geometrical distortion

geometrical distortion can be considerable and confusing in projection radiographs. the first effect is that all objects are magnified in the image. the further from the image receptor the object is placed, the greater the OID and the greater the magnification. the image size of objects, therefore, depends on their actual size and on the OID and projection direction, leading to ambiguity. this effect is illustrated in fig. 6.2(a). the three spheres a, b and c are the same size, but are projected at different sizes owing to their OIDs. furthermore, projection leads to shape distortion. in fig. 6.2(b), a tilted object is shown projected at a range of angles, illustrating the increasing degree of foreshortening as the angle increases.

1-2-3 Inverse square law

For an isotropic point source, the X ray beam intensity is inversely proportional to the square of the distance from the source. an X ray tube with its attached collimator is a good approximation to a point source for distances greater than about 50 cm from the focal spot, and obeys the inverse square law (isl) almost exactly at distances greater than this. only at low kV settings, such as those typical of mammography, does air attenuation affect the inverse square relationship. this is illustrated in fig. 3, where the air kerma per unit mas is shown over the fiD range of 50–250 cm. figure 3 also presents the calculated curve assuming the ISL.

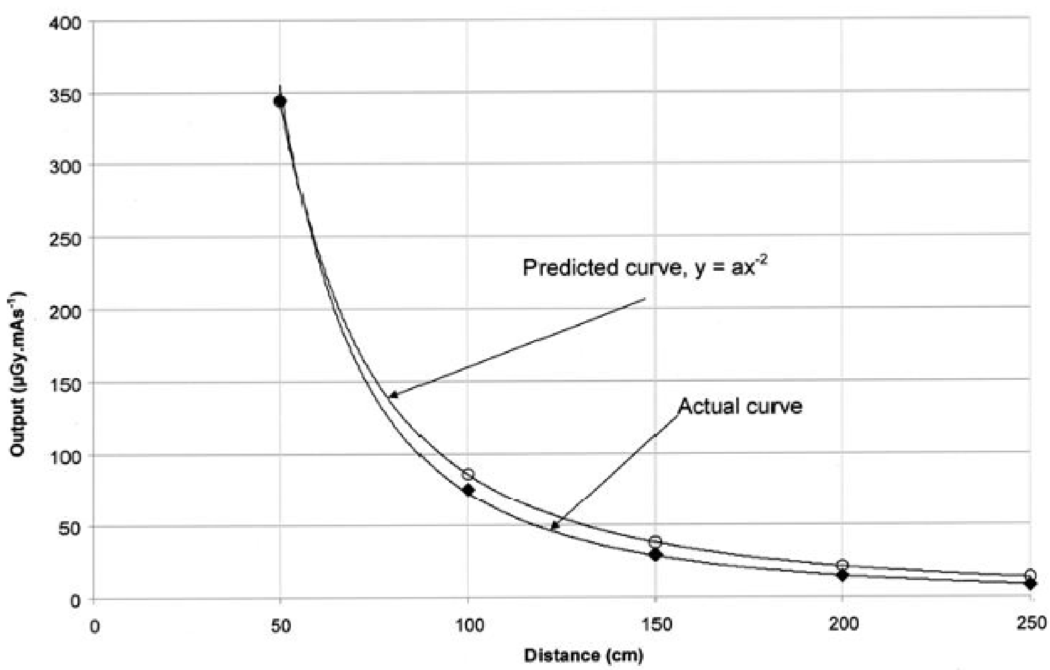

FIG. 3: Deviation from the ISL due to air attenuation for a tungsten target X ray beam with 0.5 mm Al added filtration at a voltage setting of 30 kV and no compression paddle.

in these expressions, d_{FID} is the FID and d_{FSD} is the focus to skin distance (FSD). it is easy to show that as the FID is increased, the incident air kerma (Ki) may be decreased, keeping the same kerma at the image plane; the formula for this is:

$$K_{i_2} = K_{i_1}\left(\frac{d_{FID_2}}{d_{FID_1}}\frac{d_{FSD_1}}{d_{FSD_2}}\right)^2 \quad \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots(6)$$

This relationship can be used to prevent excessive skin doses; generally, an FID of 100 cm or greater is sufficient. it does not, however, result in a similar reduction in the overall dose to the patient because an increase in the entrance surface X ray beam size is required as the FID is increased in order to prevent cut-off of the region of clinical interest. the effective dose is approximately proportional to the dose–area product. the dose reduces at longer FID according to eq. (6), but the area increases. therefore, there is little or no change in effective dose.

2- Dose

A number of effects occur when increasing the OID. there is a substantial reduction in the scatter fraction at the image receptor because the scattered rays are generally directed away from the receptor. to maintain the dose to the image receptor, an increase in the mAs, and hence the patient dose, would be required, mainly because of the loss of scatter but also because of the increase in FID owing to the ISL. owing to the reduction in scatter fraction, magnification may usually be performed without the use of an antiscatter grid. this leads to a reduction in *mAs* in proportion to the bucky factor, which is the ratio mas with a scatter reduction method divided by mas without a scatter reduction method. this factor is typically between about three and six.

3- Relationship between kV and mAs

given that the image receptor dose is proportional to mas and to kV, some simple exposure rules may be derived. firstly, it is observed that an increase in kV of 15% results in an increase in image receptor dose by a factor of two — hence the so-called '15% rule', that an increase in kV of 15% is equivalent to a doubling of the mas and a reduction by 15% is equivalent to halving the mas. furthermore, an increase in kV of 5% results in an increase in image receptor dose of 30%, leading to the '5% rule' that a 5% increase in kV is equivalent to a 30% increase in mas and a reduction of 5% in kV is equivalent to a reduction in mAs by 30%. finally, since a 15% increase in kV is about 10 kV between 60 and 80 kV, another commonly used rule is that a 10 kV increase is equivalent to doubling the mas, and a 10 kV reduction is equivalent to halving the mas. none of these rules are exact, but their use is satisfactory because of the tolerance for small exposure errors owing to the latitude of screen film systems, and because of the very wide dynamic range of digital systems


Thank you