COMPUTERIZED PATIENT RADIATION DOSE ASSESSMENT IN DIAGNOSTIC RADIOLOGY

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I. Introduction:

Reduction of radiation dose received by patients undergoing radiological procedures has become of prime importance. This emphasis has been spurred by the fear of possible biological effects including genetic damage to future populations, cancer induction and life shortening. Dose reduction in Radiology can in fact be accomplished with current available technology[1]. To this end work has proceeded in such areas as introducing faster film-screen receptor systems utilizing rare earth phosphors and in using heavy element x-ray beam filtration. Additionally, various devices have been prepared for use in x-ray scatter control as well as optimization of x-ray beam energy and the use of lower attenuating carbon fiber materials for table tops and receptor cassette fronts.

Experience has shown a general lack of both knowledge of specific patient dosage and of understanding on how various equipment operation parameters such as KVP, beam filtration, and patient related factors such as thickness and x-ray field size effect dose delivered.

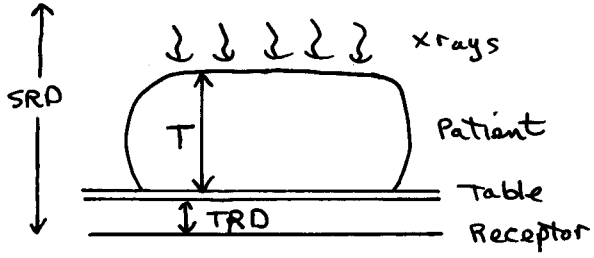
This problem has been further aggravated by the great variety of equipment features and available technologies which are available. The dose data available in the literature in general are not adequate in that they do not provide sufficient information for specific patient parameters and specific equipment utilized. That is, such data tends to be national averages for typical sized patients. There is, however, considerable dose variation due to patient thickness and radiographic techniques. One recent report shows a variation of up to 30 times within the same hospital between different rooms for the same patient.[2]

This work addresses itself to this problem. Here, we identify the factors affecting patient radiation dose, and possible methods of using digital computers to quantitate and report such dose data. From the dose models formulated insight into the full range of factors affecting patient dose and possible dose reduction schemes can be obtained. It is hoped that this work will also contribute to this formulation of more meaningful measures of dose itself as well as measurements to correlate dose with information content and its eventual optimization for a given radiological task.

II. Dose Model:

The dose model used here can be developed as follows: assume some x-ray exposure output value in air at some standardized source to receptor distance (SRD) as follows:

Figure 1



It is quite common to calibrate and express x-ray outputs (X_c) in Roentgens per mAs at a specific distance such as the SRD. From Figure 1 it is seen that the x-ray output at any point in space such as where the skin level would be located (X_s) for a patient of thickness T is given by:

$$X_s = X_c \left(\frac{SRD}{SRD - TRD - T} \right)^2 \qquad Eq.\#1$$

Where TRD is the table top receptor distance. The expression within the square is just a geometrical inverse square factor indicating expected change in air exposure at the two different points considered in space and X_c is the calibrated output of the unit at distance SRD.

Equation 1 is incomplete in three ways. First it is expressed in units of exposure (Roentgens). In addition it does not include the back scatter (BSF) provided by the patient. Finally the actual film mAs. Taking these into account, the dose to the patient's skin may then be expressed as:

$$D_s = X_c \left(\frac{SRD}{SRD - TRD - T} \right)^a \cdot F \cdot BSF \cdot mAs \qquad Eq.\#2$$

Here the factor F is the Roentgen to rad factor which is typically equal to 0.92.

Note that dose to skin here is seemingly independent of beam filtration, receptor sensitivity and losses in attenuation between patient and the receptor due to anti-scatter grids. However, these factors do enter in that they determine the necessary air exposure to obtain the desired radiographic effect (which is proportional to the film mAs). This is an extremely important

point in possible dose reporting schemes because specific knowledge of attenuation occurring in components or efficiency of receptors is not necessary. The only factors which are needed as is seen in equation 2 are the calibrated x-ray output, patient thickness, the source receptor distance (SRD), the patient thickness and the exposure parameters of mAs. In addition, note that equation 2 does depend on beam energy since BSF and X_c both depend on KVP. The factor BSF is also determined from the size of the receptor. F is approximately constant over the typical range of diagnostic x-ray beam energies. If specification of dose is desired at some depth such as the fetal depth, ovarian depth, depth to bone marrow or any specific organ, the percent depth dose, which is critically dependent on the KVP, can be incorporated as follows:

$$D_d = X_c\left(\frac{SRD}{SRD - TRD - T}\right)^2 \cdot F \cdot BSF \cdot mAs \cdot \%DD \qquad Eq.\#3$$

At present the most limiting data to satisfy equation 3 is the %DD data. In contrast to the diagnostic x-ray case exquisite data have been developed for high energy radiotherapy applications. However, data of Trout[3] is used throughout this work. An alternate approach in setting up the patient dose model is that of using tissue air ratios (TAR) which then obviates the need for the geometrical inverse square factor. TAR data is, however, very incomplete at this time. Due to their simplicity, however, future dose models will undoubtedly be designed around TAR's as more such data becomes available.

If in the present work it was found that using Trout's[3] data the variation of dose with patient depth (d) was exponential and could be described with equations as follows:

$$\%DD = A e^{-Kd} \qquad Eq.\#4$$

where K is another exponential as a function of KVP and A is a constant. These exponentials are easily modelled and programmed for rapid calculation of patient dose at any desired depth given beam exposure parameters and geometry.

Simple exponential models were also fitted for the BSF, and the x-ray outputs as a function of KVP. Finally a phase factor was derived which generalizes depth dose data to any electrical phase φ (single or three phase). This phase factor takes the form:

$$\text{Phase factor} = (BK)^{\frac{\varphi-1}{2}} \qquad Eq.\#5$$

Where B is a constant. All of the above factors were incorporated into dose model to determine organ dose for all procedures within the Radiology Department.

Bibliography:

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