Although no two projects are exactly alike, our current plans may be of use to others venturing into this field. The results of our experiences should be ready within the next year and will be available to interested parties.

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A Computer Method for Radiation Treatment Planning

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Automatic computation methods were first developed and applied to the problem of radiation therapy treatment planning by the Physics staff at Memorial Hospital and Sloan-Kettering Institute in 1954 and reported in 1955 [1]. The field of radiation from a single port was stored as a matrix in a library of punched cards, and a sorter and accounting machine were used to combine various fields for rotation, cycling and multi-port therapy. This system was in continuous routine use from then until 1961, when the equipment was replaced by a Bendix G15-D digital computer. Subsequent work by Sterling [2] followed essentially the same method of describing the radiation field as used by the Physics staff at Memorial Hospital [1], except that more powerful equipment has been used. An analytic expression for the dose distribution produced by rotation had been previously applied successfully in 1951 to treatment planning with high-energy X-rays [3].

In 1959, it appeared that the then existing system would shortly be made obsolete by certain changes in the equipment available for radiation therapy. After restriction of the number of field sizes and shapes to be punched for automatic calculation to a practical minimum for the accuracy required in therapy, a library of some 3,000,000 punched cards would still be required. This being impractical, a new method of automatic computation was looked for which would hopefully accomplish three long-desired ends: reduce data input required to a minimum; permit automatic plotting of isodose curves of the radiation field resulting from all practical combinations of ports; and permit taking into account the effects of body inhomogeneities (e.g. bones, lung and muscle).

The first attempt was simply to reduce the amount of data input required by the existing system. A special-purpose switching device was designed to be connected to and controlled by the accounting machine, permitting automatic rotation of the sampling grid for the dose matrix, reducing the data input by a factor of about 30. This would have reduced the punched-card library to manageable levels, but would not accomplish the aims of automatic plotting and calculation of effect of body inhomogeneities. Accordingly, a completely new description of the radiation field was devised which would permit these ends to be achieved.

If the field of radiation within a patient is given as $\bar{F} = f(p, q)$ where p and q are any convenient coordinates, then $\bar{F}$ is a function of the source-skin distance, S, the angle of the beam relative to the coordinate system α, the beam width w, and the beam length h. In the original Memorial-Sloan-Kettering system, separate matrices were punched for each variation in each of the above parameters. As mentioned above, rotation of the sampling grid may be accomplished with a suitably-equipped accounting machine and is a trivial problem for a computer, so that the parameter α is easily eliminated. The parameter S may be eliminated in two steps. First, a coordinate system is chosen for sampling the field consisting of the depth below the surface x, and the angular displacement from the beam centerline δ. Secondly, the effect of inverse-square-law attenuation is removed by defining a new field $\bar{M}$ such that

$$\bar{M} = \bar{F}\left[\frac{S + x}{S}\right]^2.$$

The field $\bar{M}$ is then digitized by sampling at equal increments in x and σ. For the primary beam, the matrix $\bar{M}_p$ is of rank one, being the outer product of two vectors, one of which represents attenuation and the other the transverse shape of the beam. (Beam quality is assumed invariant across the field.) Worthley and Wheatley [4] have reported that a similar relationship exists for scattered radiation in the range of half-value layers from 1 to 2 mm Cu and for source-skin distances from 40 to 100 cm; i.e., the scattered radiation field is given by

$$\bar{M}_s = u(x)\cdot v(\sigma)$$

and is also of rank one. The matrix of total radiation $\bar{M}$, being the sum of two rank one matrices, is at most of rank two and can be determined as the sum of the outer products of two factors or pairs of vectors. The data input for each calculation has then been reduced to at most four vectors instead of an entire matrix.

The actual rank of the measured total radiation field matrix may be determined by the factor analysis method of Woodbury [5]. In practice, for Cobalt-60 and 2 MV radiation, the total field matrix is very nearly of rank one. The vector representing attenuation (and now scatter as well) becomes simply the well-known tumor-air-ratio or TAR, the other vector retaining its identity as the off-center line-ratio or OCR. For Cobalt-60 or 2 MV radiation the model representing the field intensity at any point P then becomes:

$$F(P) = [\mathrm{TAR}(x)]\cdot[\mathrm{OCR}(\sigma)]\cdot\left[\frac{S}{S+x}\right]^2$$

(The angle σ is sufficiently small in practice so that the usual small-angle approximations apply.) It is then found that except for a small inaccuracy due to ignoring elongation-ratio effects the TAR vector is a function of field area alone, and the OCR vector is a function of field width alone. This reduces the total data required for a given modality to the *sum* of the field widths and areas required, rather than the *product* of the field widths and lengths required. A further source of inaccuracy should be mentioned: the TAR vector varies slightly with the source-skin distance S, and this variation is ignored. It can be easily taken into account by an empirical modification of the TAR argument x; this was not found necessary. Inaccuracies due to effects of curvature and inhomogeneities, for which the method described can correct, were considered much more important.

This model may be used to account for inhomogeneities to a first-order approximation, i.e. to correct for differential absorption. Since in the Cobalt-60 and 2 MV range absorption by body materials is almost exclusively Compton, the argument for TAR may be modified:

$$\mathrm{TAR} = \mathrm{TAR}(\Sigma \rho t)$$

where the density ρ is expressed in terms of electrons per cubic centimeter as compared to water as a standard.

Referring to Figure 1, the model for rotation becomes:

$$\mathrm{Dose}(P) = \int [\mathrm{TAR}(l(\theta+\beta) - \rho\cos\beta)] \cdot \left[\mathrm{OCR}\left(\frac{\rho\sin\beta}{F-\rho\cos\beta}\right)\right]\cdot\left[\frac{F}{F-\rho\cos\beta}\right]^2 d\beta$$

Cycling and complete rotation are simply calculated by adjusting the limits of integration; for multi-port therapy, summation is used. By suitable conversion of the coordinate system actually used to those employed by the matrix $\bar{M}$, any desired treatment may be calculated. Curvature may be accounted for by either an iterative or stepping method. If polar coordinates are used, as shown in Figure 1, the

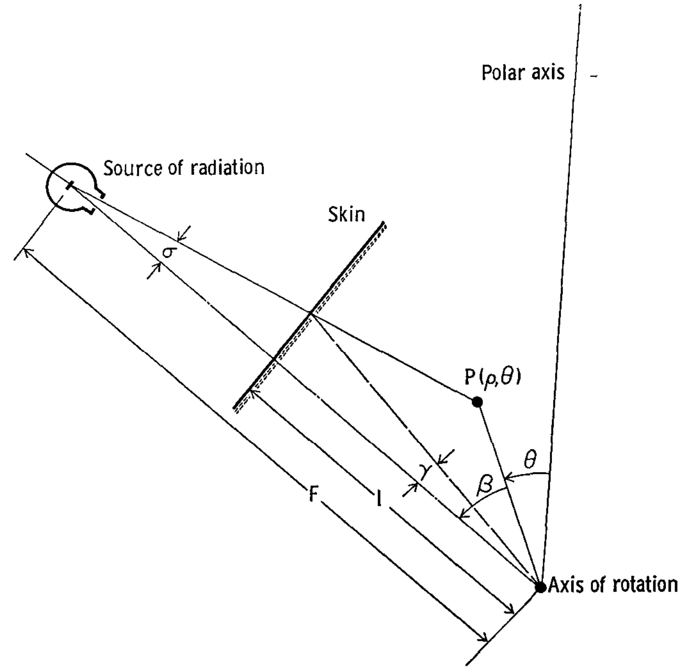

FIG. 1. Typical geometry for rotation therapy calculation

transcendental equation

$$l(\theta+\beta-\gamma)\sin(\gamma+\sigma) = \rho\sin(\beta+\sigma)$$

must be solved, l being interpolated from the stored patient contour. The quickest solution is:

$$\sin(\gamma_{i+1}+\sigma) = \frac{\rho\sin(\beta+\sigma)}{l(\theta+\beta-\gamma_i)}$$

but the convergence is uncertain if the dose point is very near the skin, and suitable precautions must be taken.

The rotation model given above is suitable for an analog computer and was in fact originally so programmed; but the greater power and flexibility of the digital computer for general medical work caused abandonment of the analog-computer work. Flow charts for the program are available on request.

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