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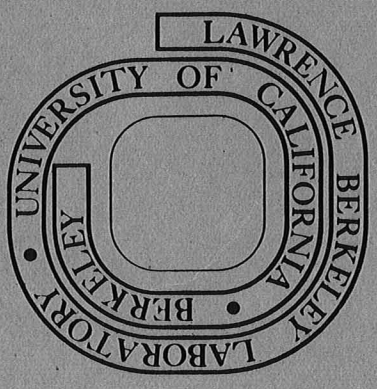

A 3-DIMENSIONAL BEAM SCANNING SYSTEM FOR PARTICLE RADIATION THERAPY*

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Summary

In radiation therapy treatment volumes up to several liters have to be irradiated. Today's charged particle programs use ridge filters, scattering foils, occluding rings collimators and boluses to shape the dose distribution.[1] An alternative approach, scanning of a small diameter beam, is analyzed and tentative systems specifications are derived. Critical components are scheduled for fabrication and testing at LBL.

Introduction

Apart from biological advantages whose detailed investigation constitutes at present a substantial research effort heavy charged particle beams have the physical advantage of spatially extremely well defined dose distributions. Their successful application in a wide range of therapeutic situations requires means to irradiate volumes as large as 30cm x 30cm x 15cm. Prescribed dose levels should be attained within a few precent, irradiation of healthy tissue minimized and the distortion of dose distributions caused by tissue heterogeneities should be compensated for. The time for one treatment should not exceed a few minutes; ideally about 1 minute is aimed at. 3-dimensional scanning seems to meet these requirements. The basic components of a scanning system are two magnets sweeping the beam over the treatment area, a device or procedure to change beam energy quickly, a device to control beam intensity and a computer control system.

Analysis of Procedure

Spatial Characteristics

We briefly examine the relation between the prescribed dose distribution D(X,Y,Z) and the sweep pattern. From this criteria for the step sizes will be derived. We assume that over the duration of one treatment (~60s) the detailed time structure of dose accumulation is biologically immaterial. The scanning pattern is then described by a distribution function I(X,Y,Z):

$$\int_{-\infty}^{x}\int_{-\infty}^{y}\int_{o}^{z} I(X_o,Y_o,Z_o)\, dX_o dY_o dZ_o = N(X,Y,Z) \qquad (1)$$

where N(X,Y,Z) is proportional to the total number of particles delivered with beam centroid positions $X_o \leq X, Y_o \leq Y$ and range $Z_o \leq Z$. We describe the contribution to the dose at (X,Y,Z) from a beam with centroid coordinates (X_o,Y_o) and range Z_o by a function $g(X,X_o,Y,Y_o,Z,Z_o)$. The relation between I and D is then expressed as a Fredholm equation of the first kind:

$$D(X,Y,Z) = \int_{-\infty}^{+\infty}\int_{-\infty}^{+\infty}\int_{z1}^{z2} I(X_o,Y_o,Z_o)\, g(X,X_o,Y,Y_o,Z,Z_o)\, dX_o dY_o dZ_o. \qquad (2)$$

Exact solutions of this deconvolution problem will have to be found numerically and $g(X,X_o,Y,Y_o,Z,Z_o)$ has to be determined first by solving a "transport" or "Boltzmann"- type integro-differential equation.

To define systems parameters a simpler approach suffices. We assume a homogeneous medium, neglect secondary beam components resulting from nuclear interactions and assume a beam profile indepedent of Z. Then we obtain:

$$g_x(X-X_o,Y-Y_o,Z-Z_o) = g_x(X-X_o).g_y(Y-Y_o).g_z(Z-Z_o). \qquad (3)$$

This allows a simple solution of equation(2) in terms of Fourier transforms:

$$\bar{D}(k,\ell,m) = \bar{I}(k,\ell,m).\ \bar{g}_x(k).\ \bar{g}_x(\ell).\ \bar{g}_z(m) \qquad (4)$$

where the bar denotes the Fourier transform. The (spatial) frequency range of D, and therefore attainable resolution, is obviously limited by the finite beam spot size and longitudinal extent of the Bragg peak. Also a finite step scanning pattern, resulting in a δ-function like I(X,Y,Z) yields very smooth dose distributions (variations < 1%) if the step size does not exceed ~ one beam half-width. Lower limits for beam spot sizes and therefore useful step sizes are determined by multiple scattering and are of the order of 2 to 3mm for ^{12}C and 5 to 7mm for protons for ranges in tissue between 20cm and 30cm.

Sweep Velocity Requirements

In the following required magnet rise times are derived. We envisage a system with a fast scan in X-direction at constant (Y,Z), a slower scan in the Y-direction and the slowest change in Z, i.e. energy. The treatment volume is thought to be subdivided in cubical elementary volumes or "cells" whose length L_c corresponds to the characteristic beam dimensions. For a given treatment volume V, treatment time T and macroscopic duty cycle η of the accelerator we obtain for the scanning velocity in X-direction, v_x, and the corresponding time Δt_x:

$$v_x = (\eta T)^{-1}\, V/L_c^2 \quad (3) \quad \text{and} \quad \Delta t_x = \eta T\, L_x L_c^2/V. \qquad (4)$$

There exists a minimum time $\Delta t_{x,min}$ independent of treatment volume corresponding to $L_{x,min}$, the minimum

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length of the X-scan compatible with a given volume:

$$\Delta t_{x,min} = \eta T \, L_c^2/(L_{y,max} \cdot L_{g,max}) \qquad (5)$$

$$L_{x,min} = V/(L_{y,max} \cdot L_{g,max}) \qquad (6)$$

If a finite-size step pattern is used an efficient system requires a sweep speed v_y in Y-direction given by:

$$v_y = k.\frac{L_c}{\Delta t_{x,min}} \qquad k \sim 5 \qquad (7)$$

The time interval required to complete a 2-dimensional scan at a given energy is of the order $\eta T \, L_c/L_g$ and energy changes must be performed in a fraction thereof or be synchronized with beam-off periods, e.g. between synchrotron pulses.

Intensity Control

In order to achieve the desired dose distribution energy deposition at each scan position has to be controlled by controlling beam intensity and/or adjusting the scanning speed. Our design uses a finite-step X-scan, accumulating dose at constant X, then stepping to the next position with a sweep speed of 100 ms^{-1} for a 400 MeV/amu Carbon beam. A fast (a few μs) beam switch allows the beam to be turned off during the step if required. The design sweep speed while stepping is 5 times greater than the value V_x from equation (5) for L_c = 4mm, ηT = 30s and V = 10 liters. For a 30cm scan this corresponds to going from $-B_{max}$ to $+B_{max}$ in 3ms.

System Design

Scanning Magnets and Power Supplies

Optimum magnet parameters depend on the properties of the overall beam delivery system in which they are incorporated. In horizontal beams they are easily incorporated in the final drift length which can be fairly long. In the vertical beams or in the isocentric systems considered for therapy overall economics may dictate a location further upstream. We estimate the power requirements for the first scanning magnet. The magnetic field is:

$$B_{max} = \frac{\Theta(B\rho)}{L_m} = \frac{L_x(B\rho)}{2\,T_{12} \cdot L_m} \qquad (8)$$

where T is the transfer matrix from magnet center to target, L_m the magnet length and L_x the length of the scan. The stored energy is:

$$U_s = 4.\frac{10^7}{8\pi}\,\alpha\,\frac{\varepsilon\sqrt{\beta_x}\,\sqrt{\beta_y}}{L_m} \cdot \left(\frac{L_x}{2}\right)^2 \frac{(B_\rho)^2}{T_{12}^2} \qquad (9)$$

where $\pi\,\varepsilon_x = \pi\varepsilon_y = \pi\varepsilon$ is the beam emittance, β_x, β_y are the beta functions at the magnet and the factor α takes into account that the actual magnetic field volume is somewhat larger than the beam aperture. For a spot size $X_f = \sqrt{\varepsilon}\,\sqrt{\beta_f}$ we obtain $\beta_{x,min} = T_{12}^2/\beta_f$, $\beta_{y,min} = T_{34}^2/\beta_f$.

Equation (9) becomes:

$$U_s = \alpha\,\frac{10^7}{2\pi} \cdot \frac{\varepsilon^2 (B\rho)^2}{L_m} \left(\frac{L_x}{2\,x_f}\right)^2 \cdot \frac{T_{34}}{T_{12}} \qquad (10)$$

For α = 1.5, $\varepsilon = 10^{-5}$m, x_f = 3mm, L_x = 30cm, L_m = 1m and $B\rho = 6.5\ T_m$ we obtain:

$$U_{s,min} \simeq 25.\frac{T_{34}}{T_{12}} \ (J)$$

and with a risetime of 1.5ms (0 to B_{max}) the required peak power is :

$$P_{peak} \simeq 35.\frac{T_{34}}{T_{12}}$$

This approximate analysis of power requirements is valid if:

$$T_{12}^2 >> \frac{L_m \cdot L_x \cdot X_f}{4\varepsilon}$$

Also if T_{34} is very small it's maximum value over the magnet length must be used. At present a ferrite core scanning magnet is being built at LBL. Apertures are slightly larger than minimally required in order to have some flexibility in ion optical arrangements. Typical parameters are: L_m = 1m, aperture = 100mm x 40min, B_{max} = 0.22T and U_s = 100J.

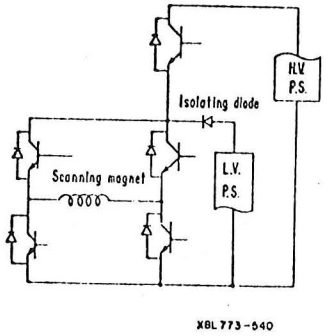

Fig. 1 Fast scanning magnet Power supply schematic.

In all investigated layouts the stored energy in the second magnet is larger, ~1.0 to ~ 3.5 kJ, due to increased aperture requirements and this magnet is therefore the logical choice for the slow scan.

The fast magnet is powered by four transistor actuators in a bridge circuit allowing a constant current of either polarity by regulating the appropriate actuators. Power for the holding current is provided from a low voltage power supply connected through an isolating diode. A fifth actuator switches

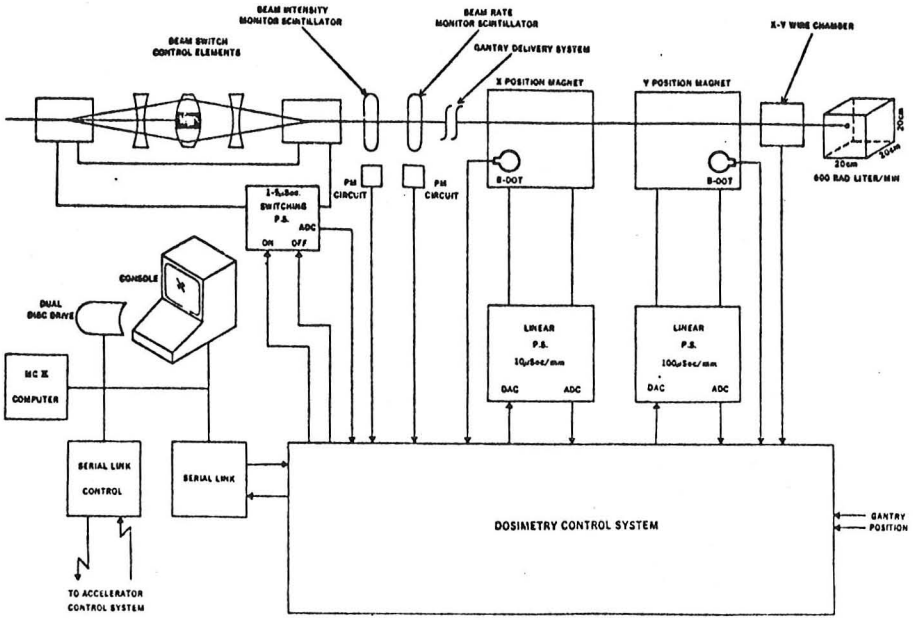

Fig. 2
Control System
Schematic.

a 100 V power supply to the bridge and by turning on hard the appropriate actuators in the bridge the magnet is driven to the new holding current. The fifth actuator is switched off and the regulating mode resumes. (Fig. 1) Work on a test power supply has started.

For the slow magnet an analogous approach substituting SCR's for the transitor actuators is planned.

Beam on/off Switch

The beam switch simple consists of a magnet capable of sweeping the beam by one to two beam diameters over a matching collimator in a few μs. Preliminary designs envisage a 1 to 2m long ferrite core, one turn magnet operating at fields of less than 10m T.

Energy Modulation

It seems entirely feasible that a synchrotron designed for medical applications could perform energy changes on a pulse to pulse basis. In the interest of simple operation however, or for installation at existing facilities such as the LBL Bevalac, a variable thickness degrader appears more attractive. Such a degrader must be located after the last beam transport elements very close to the patient since it introduces a substantial increase in emittance. For protons, for example, even if a low Z degrader such as Be or C is used, emittance in each plane increases by a factor of approximately 10 to 50 and an energy spread of a few MeV is introduced for the required range adjustments. A device similar to the LASL range shifter[2] seems ideally suited.

Monitoring and Computer Control

Fig. (2) depicts schematically a tentative design of the control system. Two scintillator and photomultiplier circuits are used, one for monitoring dose accumulation and one for continuous rate monitoring allowing interruption of the process if the beam intensity exceeds as preset level. Scanning magnet fields are measured by B-coils with a multi-wire proportional chamber serving for position verification With only a few hundred μs allotted for each scan position real time computer control will be difficult if a precision of a few percent is aimed at. Actual control and data acquistion will be performed in hardware (dosimetry control system) with the computer interacting on a plane to plane basis, i.e. when an energy change occurs. The computer, e.g. a ModComp MCII, will be equipped with a control console, a disk drive containing the dose profiles and two high speed serial links, one to the accelerator control computer and one to the dosimetry control system.

In a plane Z_k each cell is characterized by the set (D_t, X_i, Y_j, B_z), where D_t is the total specified particle count, X_i, Y_j it's (X,Y) - coordinates and B_z a modifier used to calculate the cell memory address from the scanning magnet fields B_x, B_y for each energy.

In operation the Z_k, dose profile is downloaded in one half of the local memory in the dosimetry control system. The beam is positioned at X_{kmax}, Y_{jmax} and the cell is exposed until the desired dose is obtained (or exceeded). Address verification and dose comparison is performed on the basis of a $1\mu s$ cycle. Then the actual dose is recorded and the beam is moved to X_{kmax-1}, etc. When all X_i, Y_{jmax} are satisfied a step to Y_{jmax-1} occurs. While the plane Z_k is exposed, actual recorded doses at Z_{k-1} are transmitted back to the disk and desired dose values for Z_{k+1} are loaded into the local memory.

Conclusion

It seems feasible to construct a beam delivery system which allows therapeutic irradiations to be performed with a precision limited only by the physical characteristics of particle beams and their interaction with matter. Critical subsystems, fast scanning magnet, power supply and parts of the control system are under construction at LBL. A meaningful application of this technique requires of course diagnostic procedures of equal precision. Such techniques are becoming available however with 3-dimensional reconstruction procedures (i.e. CAT scans) or the use of auto-radioactive beams.[3]

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