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PION AND HEAVY ION RADIOTHERAPY

Pre-Clinical and Clinical Studies

Editor: L. E. Skarsgard

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Pion and Heavy Ion Radiotherapy:

Pre-Clinical and Clinical Studies

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PION AND HEAVY ION RADIOTHERAPY:
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CRITERIA AND TECHNIQUES FOR THREE-DIMENSIONAL TREATMENT PLANNING WITH PIONS

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INTRODUCTION

The ability to predict a pion dose distribution in a patient is a major objective of the clinical trials at LAMPF. Accurate predictions are essential for evaluation of pion therapy. But accuracy must be in the context of clinical utility. That is, reasonable approximations must be made in calculational methods so that treatment planning can proceed in a timely and efficient manner. We present here a few of the techniques and current developments used to achieve that objective.

METHOD OF DOSE CALCULATION

The treatment-planning program PIPLAN calculates a dose distribution by summing the contributions of individual pencil beams as they pass through various clinical appliances and patient anatomy. One-dimensional depth dose distributions for pencil beams of each type of incident particle and for each of their dose components are precalculated analytically in water and include the effects of inflight interactions, straggling, and decay. Additionally, a two-dimensional distribution is precalculated for long range neutrons that are emitted by inflight and stopping pion interactions. Secondary muons from pion decay are treated as a composite psuedo-particle with its own dose distribution. Primary electron dose is obtained from fits to high-energy electron depth-dose curves. Long-range electron dose from muon decay is obtained from a separate model during the dose accumulation process.

For a given pencil beam, dose is distributed in depth as a function of water-equivalent range along the incident trajectory and then further distributed radially as a function of multiple Coulomb scattering. A dose calculation then entails accumulating at each point of interest the amount of energy from each incident pencil beam and then dividing by the local mass density. To model the complex, non-analytic, phase space of treatment beams, measurements of individual trajectories or rays are used as input to the

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calculation. This introduces a Monte Carlo aspect into the calculation and associated statistical uncertainties. However, the otherwise analytic nature of the calculation and distributed dose for each pencil beam yields a result comparable to Monte Carlo in much less time.

PIPLAN calculations are initially verified with measurements of unmodulated beams in water. Figure 1 shows a typical comparison after slight adjustments in the beam momentum and background. However, these adjustments are within experimental uncertainties, are only beam-tune dependent, and are then constant for all subsequent calculations. PIPLAN verification then consists of comparing predicted and measured dose distributions for various beam-modulation functions and geometries. Predictions typically match measurements to better than four percent of maximum, as shown in Figure 2. An inability to measure the entire beam phase-space passing through the primary monitor chamber presently requires that normalized comparisons be made.

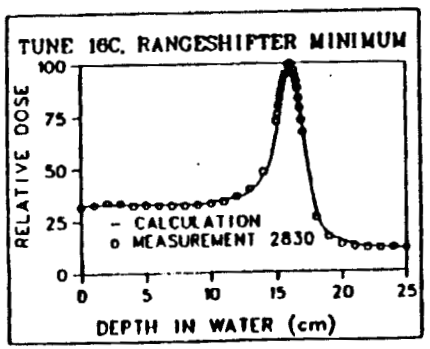

Fig. 1. Comparison of calculation and measurement of depth-dose curves for an unmodulated, medium-energy, large-field beam.

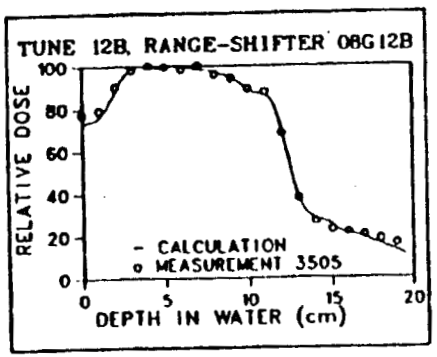

Fig. 2. Comparison of calculation and measurement of depth-dose curves for a modulated, low-energy, medium-field beam.

CT DATA

The value of CT data in radiotherapy in general is widely accepted. In particular, it is used for at least three different purposes: diagnosis, treatment planning calculations, and evaluation of dose distributions on a CT image background. For the latter two purposes it is possible to significantly reduce the bulk of CT data without significant loss of calculational accuracy or visual resolution. The corresponding reduction in processing time, data storage space, and data transfer time between machines can be substantial, especia for computers capable of 8-bit byte processing.

TABLE 1

RESULTS OF REDUCED CT NUMERICAL PRECISION AND SPATIAL RESOLUTION

SITE	TEST CASES	AVERAGE INTEGRAL CHANGES (mm)	AVERAGE INCREASE CHANGES (mm)
SINUSES	4/9	-0.14 (0.40)	0.0 (0.08)
	-1,+1	-0.07 (0.38)	0.0 (0.08)
	BOTH	-0.07 (0.37)	0.0 (0.08)
LUNG	4/9	0.0 (0.05)	0.0 (0.05)
	-1,+1	-0.02 (0.23)	0.0 (0.08)
	BOTH	-0.02 (0.23)	0.0 (0.08)
PELVIS	4/9	0.0 (0.06)	0.0 (0.06)
	-1,+1	0.0 (0.10)	0.0 (0.07)
	BOTH	0.01 (0.10)	0.0 (0.07)

Treatment-planning calculations use CT data for line integrals to trans particles or modify dose distributions based on effective range St integration is an averaging process, numerical and spatial resolution dim in importance. Table 1 summarizes the results of a detailed study to dete practical limits for reduced CT numerical precision and spatial resolution u three different CT slices. The procedure was to calculate line integrals wit 1 mm step size vertically for each slice and tabulate the integrals on a sq centimeter grid. This was done for three slices with full resolution (12 pixel, about 1 mm x 1 mm x 1 cm) and for various test cases. "4/9" indicat rounding procedure to reduce the numerical precision to eight bits and allow CT pixel to be stored in one byte. "-1,+1" indicates three pixel lat averaging for each integrand. When considered with the integration, this analogous to a thirty-pixel areal average between vertical grid poi Finally, the last test case used both reduced resolution techniques. changes from the full-resolution cases are shown with standard deviation parentheses and assumes a linear relation between CT number and stopping po In fact, for heavy charged particles in material more dense than water,

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values will be even smaller. The average integral change is the average change between a final base case integral and a final test case integral. For the average increase change, the average is over the changes in the increase of all the integrals from one centimeter to the next in depth. Results such as these show that accurate calculations can be retained with at least a factor of eight reduction in the bulk of CT data through four pixel spatial averaging and eight bit numerical resolution.

Resolution reduction of CT image data, as opposed to CT treatment-planning data, is subject to more subjective analysis. Visual comparisons of various images has led to the conclusion that four-pixel averaging and eight-bit precision also retains sufficient resolution and detail in a CT image for accurate evaluation of superimposed dose distributions.

Finally, we note that while these techniques preclude regeneration of a diagnostic quality image, they do allow random access of individual CT data elements. This means that treatment-planning calculations and displays can proceed equally well at any orientation relative to the original CT slice.

MULTIPLE SCATTERING

A major objective of treatment-planning calculations is the accurate prediction of transverse dose distributions as a function of depth and their dependence upon multiple scattering in external appliances and the heterogeneous anatomy of a patient. PIPLAN uses a recursive Gaussian model which is energy, material, and geometry dependent. The transverse distribution at any point is the result of folding together Gaussian distributions from multiple scattering at many points upstream. Although the angular distribution from multiple scattering in upstream regions is used to obtain the radial distribution downstream, in this model the angular and spatial distributions are uncoupled. That is, each scattering region assumes all incident particles are still parallel to the initial trajectory but are spatially distributed about it according to upstream scattering. This feature allows a very rapid recursive expression, but only works well for ideally thin scattering regions. To model thick targets, a modifying function is used which effectively requires that the transverse, spatial standard-deviation from a single thick target equals that obtained from many thin targets of the same total thickness. This function essentially compensates for the uncoupling of the angular and spatial distributions and allows large distances to be traversed in a single cycle, such as air gaps between the range modulator and the patient. The function includes

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two constants which determine the effective energy-loss of a particle. Th result is a recursive model that is both fast and accurate. The deg accuracy is illustrated for pions in Figure 3 for an unmodulated beam in A typical step size of 0.5 cm was used in this calculation.

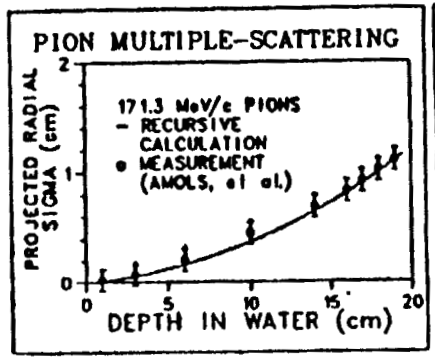

Fig. 3. Comparison of calculation and measurement of projected standard deviation for Gaussian transverse distribution as a function of depth.

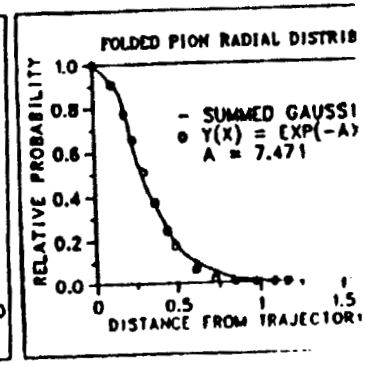

Fig. 4. Comparison of Gaussian weighted and summed Gaussian tra distributions obtained at 5 cm f for a 3-cm range modulation and air-gap.

When beams are range modulated, the transverse distributions of a give due to different upstream scattering geometries must be added with obtained from the range modulation function. Since the sum of Gaussians conveniently analytic, the final transverse distribution can be obtai tracking each ray repeatedly as a function of the range-shifter thickne sampling the modulation function for each ray. The first is expensive and the second can introduce significant statistical uncertainties in the spread peak unless a great number of rays are used. We are c investigating the feasibility of tabulating for each modulation functi of parameters at each depth along a trajectory which describe the distributions. We find that in most geometries a Gaussian distributio quite well, as shown in Figure 4. For very large modulation functio exponential function gives only a slightly better fit than a Gaussian. model is successful, then the precalculated one dimensional dose distri can be folded through the modulation function and each incident distribute dose radially and in depth as if it had experienced full modu

PHASE SPACE SMOOTHING

Another important aspect in predicting transverse dose distributions is accurately representing the phase space of treatment beams, which at LAMPF are non-parallel and contain spatially non-uniform ratios of pions, muons, and electrons. To accurately sample and transport a sufficient number of particles in a reasonable time, two techniques are used.

The first involves measuring a two-dimensional, transverse dose distribution in air and also sampling the true phase space by measuring about 250,000 individual particle trajectories. Then from the particle data, a sub-sample is selected which, by independent calculation, reproduces the dose distribution in air. The particle randomness of the original sample is retained in the sub-sample, but the statistical uncertainty in the transverse dose distribution is reduced to that of the dose measurement. The number of particles in the sub-sample is selected to achieve a desired maximum, statistical uncertainty in PIPLAN calculations. This in turn involves the second smoothing techique.

The relative statistical uncertainty at some point, due to spatially random, overlapping, radial distributions, is inversely proportional to the standard deviation of the distributions and to the square-root of the fluence. Thus for a given uncertainty, there is a direct relationship between multiple scattering and the sub-sample fluence. When the fluence is also given, the uncertainty is determined only by the degree of distribution overlap. Near the patient surface, multiple scattering overlap is small and a minimum radial standard deviation is used to obtain about three percent statistical uncertainty. Normal multiple scattering then yields equal or less uncertainty at depth. Figure 5 compares calculated transverse air scans with the ion chamber measurements for a radial sigma of 0.5 cm, which is the usual minimum. At depth, the multiple scattering sigma exceeds the minimum and the statistical uncertainty is negligible, as shown in Figure 6 for a sigma of 1.2 cm. These results were obtained for a central-axis fluence of only 43 particles per square centimeter. In Figure 7 are shown the transverse scans in a water phantom at mid-peak in a 10-cm spread peak and a central-axis fluence of 50 particles per square centimeter. These curves show essentially no statistical fluctuations and perhaps a small systematic coordinate misalignment.

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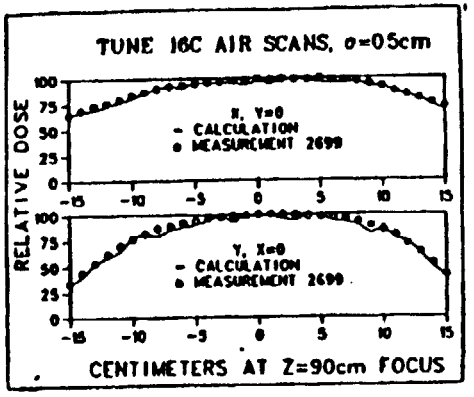

Fig. 5. Comparison of calcul. and measurement of air dose o. major transverse axes with mi mum pencil-beam sigma.

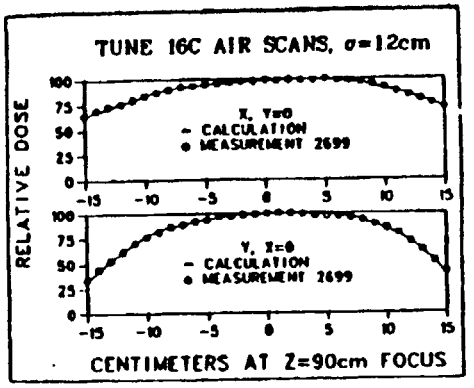

Fig. 6. Comparison of calcu and measurement of air dose major transverse axes with t pencil-beam sigma at depth.

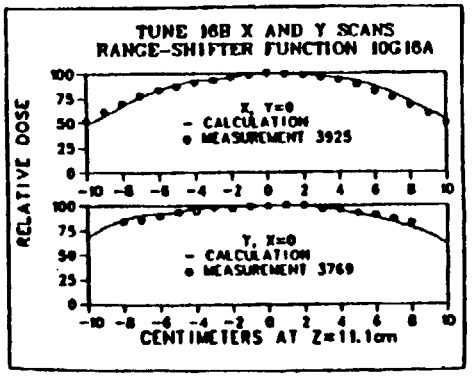

Fig. 7. Comparison of calc and measurement of water do major transverse axes at de modulated beam.

CURRENT PROBLEMS

Collimators: Three-dimensional methods of calculation, such as Monte Carlo or three-dimensional ray tracing, must accurately predict dose distributions in complex geometries and materials. This is particularly important in the regions of gross inhomogeneities, such as for a beam collimator. Here PIPLAN exhibits a problem as illustrated in Figure 8. On the left are the depth dose curves for a static rangeshifter. On the right a collimator has also been used and a clear discrepancy between calculation and measurement exists in the entrance region. Preliminary models which attempt to account for this entrance dose with neutrons from pion capture in the carrobend collimator have not been satisfactory. As shown in Figure 9, when this beam is modulated, the discrepancy accumulates. Monte Carlo calculations are proceeding to aid our understanding of collimator physics.

Dynamic Beam Bolus: Except for strictly parallel beams, bolus design for moving beams and/or patients presents an interesting problem, as schematically illustrated in Figure 10. In the upper left is represented the superposition of a given beam at three different focal points in a target volume, but ignoring the inhomogeneity. Each point in such a composite can be thought of as a point source of beam particles. For bolus, the points of interest are at the patient surface, as represented at the upper right. The bolus at that point must be such that the inhomogeneities are accounted for and that the optimum dose distribution is achieved in the target volume. With unlimited computer resources and time, one could, in principle, perform iterative dose calculations to arrive at the optimum bolus and dose distribution simultaneously. In practice one must settle for something less. Using ray tracing techniques, such as in PIPLAN, one can determine a bolus-thickness distribution for rays passing through the point in question, as indicated in the lower half of Figure 10. Rays which can be made to stop in the target volume give the distribution on the left. Of course some rays may always miss the target volume and collimation or very thick bolus should be used for them, which gives the distribution on the right. The relative number of target-volume stops and misses determine if bolus or collimator is best at a given point. If bolus is used, then the bolus-thickness distribution gives, to first order, the depth-dose distribution under that point due to particles passing through that point. This can be used to select an actual thickness from the distribution depending upon the location of nearby critical structures or various other criteria. Since bolus itself can be very structured and act as a superficial inhomogeneity in highly non-parallel beams, it may be worthwhile to iterate at least once to refine the bolus shape.

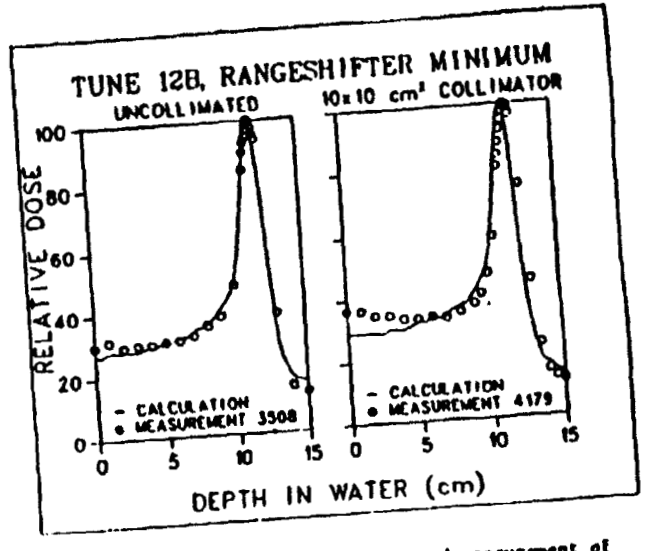

Fig. 8. Comparison of calculation and measurement of water depth-dose curves without and with collimator.

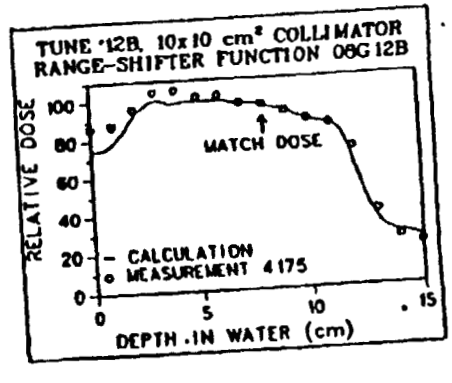

Fig. 9. Comparison of calculation and measurement of depth-dose curve for collimated and modulated low-energy, medium-field beam.

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In any case once the bolus is designed, calculations can proceed and the results evaluated.

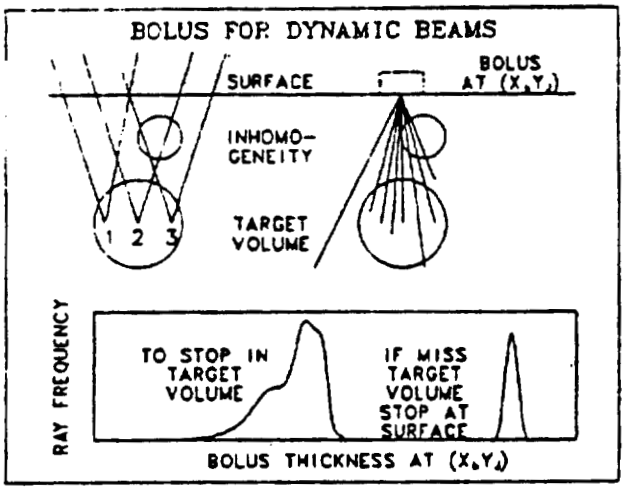

Fig. 10. For overlapping non-parallel beams, bolus-thickness distributions are obtained at each point on the patient surface. The thickness which optimizes the dose distribution is not obvious.

ACKNOWLEDGMENT

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REFERENCE

1. Amole HI et al (1981) Multiple Scattering Distributions for Therapeutic Pion Beams. Phys Med Biol 26(2): 277-289.

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