A NEGATIVE PION BEAM TRANSPORT CHANNEL FOR RADIOBIOLOGY AND RADIATION THERAPY AT TRIUMF

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Abstract

A five quadrupole, two bending magnet negative pion beam transport channel has been designed to deliver a physical dose of 0.2 rad/sec into a volume of 10 cm x 10 cm x 8 cm thick located at a mean depth of 24 cm. The dose as a function of depth will be varied by a momentum-defining slit which will travel during the irradiation; a second slit in the system will be used to give a sharp lateral cut-off in the dose. Beam size will be varied by appropriately choosing the distances from the second bend to the last quadrupole and from the last quadrupole to the irradiation location. The changes in beam properties due to second-order terms and their correction by sextupole magnets and pole edge curvatures will be described.

Discussion

There is a great deal of interest in the potentialities of negative pi-mesons in the treatment of deep-seated localized malignancies, because a beam of such particles should give a distribution of dose with depth that is significantly better than any other type of external radiation beam.[1,2] If one can control the momentum distribution of the pions reaching the patient, much better dose localization to the tumor volume is possible than with the conventional x-ray and γ-ray therapy. There are two primary reasons for this. Firstly, the range of a pion in an absorber is a well-defined function of particle momentum, and, according to the Bragg curve for loss of kinetic energy by a charged particle, the linear energy transfer is highest at the end of the pion's range (15 MeV in the last cm). Consequently, biological damage per unit volume is highest in the pion stopping volume. Secondly, when the pions stop in the absorber, the fission fragments from the pion "stars" deposit approximately 25 MeV within 1 cm of the stopping point. The combination of these two factors make negative pions particularly well suited to localization of radiation dose.

The transmission of a pion beam suitable for radiation therapy imposes diverse design criteria on the beam transport system. One principal requirement is that of an adequate beam intensity. Pion beams have not as yet been used for radiation therapy because existing beams have dose rates far too low to allow therapeutic doses in a reasonable treatment time. The minimum dose rate for the TRIUMF biomedical pion channel was specified to be 0.2 rad/sec to a treatment volume of 800 cm^3 (10 x 10 cm area, 8 cm thickness). The implied pion intensity is 2.5 x 10^8 pions per second at the irradiation location.

With a 20 gm/cm^2 carbon target placed in TRIUMF's 100μA, 500 MeV proton beam, a sufficient flux of pions with mean energy 80 MeV (range of 20 cm in water) will be transmitted through a transport system with an acceptance of 0.19 MeV-sr. A more intense pion beam will be available if the TRIUMF accelerator is operated at 380μA, 450 MeV. The shortest channel length consistent with shielding requirements around the production target is approximately 8.0 meters. The channel must be kept as short as possible: the survival of 80 MeV pions after an 8 m drift length is only 43%, decreasing to 33% at 50 MeV. In addition, the drift distance after the last bend should be minimized to keep the muon contamination down.

It is proposed to take the pions at an angle of 30° above the forward direction of the proton beam bend the pions through two successive co-planar 45° bends so that the pion beam emerges into the irradiation laboratory horizontally, perpendicular to the proton beam and some 8 feet above it. This angle was chosen both because it is the smallest forward angle allowable by the space available around the target, and because in this configuration the target presents its narrow dimension (0.4 cm) to the plane of dispersion of the bending magnets. This second consideration allows a momentum resolution of 0.5%. Range straggling affects the stopping distribution of the pions to the extent that, for the purposes of radiation therapy, a resolution of 1% would be adequate.

The uniformity of the radiation dose, both in cross-section and in depth, are also important considerations. The pion beam should be constant over a cross-sectional area to better than ±5%. This requirement is easily fulfilled for x-ray and γ-ray beams by the use of compensating filters, but such a solution is not possible with a pion beam. One solution is to place divergence-limiting apertures in the path of the pion beam to make certain that each point in the irradiated field sees the same angular range of particles from the target. Since the pion production is nearly isotropic over the angular range accepted by the channel, this should assure first order beam uniformity. Another solution is to form the pion beam into a narrow strip and scan the beam across the desired area by moving the patient across in front of the beam. Probably both modes of operation will be used with this channel.

Achieving a uniform distribution of dose with depth is another sticky problem. For one thing, a uniform distribution of stopping pions with depth does not give a uniform dose distribution because all of the pions that stop at the deep end of the tumor contribute to the dose at the shallow region, but the converse is not true. For another, the biological effect will not be uniform with depth even if the ionizing energy is deposited uniformly, so that the momentum distribution that will give a uniform biological effect with depth is not well known. The solution to this problem at TRIUMF is to make the momentum-limiting apertures in the system variable during the irradiation time, so that the momentum spectrum integrated over the whole treatment period can have any desired shape. However, in the beam scanning mode mentioned previously, the momentum slits should travel through their pre-programmed path in a time that is short compared with the time the beam traverses the treated area.

The layout of the final channel is shown in Fig. 1. It consists of the two 45° uniform field bending magnets, five quadrupole focussing magnets, and two sextupole magnets. The system between the entrance of B1 and the exit of B2 is symmetric about the center of quadrupole Q3, which is also the location of the momentum focal plane. The momentum-limiting slits (digitally controlled through stepping motors) have to travel, at most, 12 cm in a period of several seconds. The maximum axial energy transmitted will be 110 MeV, which gives a pion range in water of over 30 cm.

The beam size should be variable over as large a range as possible. By varying the position and field gradient of the last quadrupole magnet and the position of the irradiation location, there is sufficient flexibility to allow beams varying in size between 3 x 5 and 10 x 10. Smaller beams will require collimation by a beam-limiting aperture before B2, and larger treatment areas will require beam scanning.

The calculations of the optical properties of this channel are all compatible with the program TRANSPORT,[3] and all second order beam transfer matrix elements were calculated using that program. On the basis of these calculations, it was found necessary to correct for several of the second order effects: rotation of the momentum focal plane, changes in beam uniformity due to geometric aberrations, and variation of beam size with energy at the irradiation location. The first and third of these are both caused by the second order coefficient T_{126}. Because of the large divergence accepted from the target (±50 mrad) and the large momentum acceptance (±10%), this aberration can cause a strong dependence of X-plane beam size on particle momentum. If uncorrected, the momentum focal plane inside Q3 would be rotated by nearly 60° from the perpendicular to the optic axis. At this angle, even 1% resolution would be difficult to achieve. The variation of beam size with energy would be most noticeable when using a beam that is very narrow in the X direction. If uncorrected, for example, the beam might be 5 cm wide at a depth of 12 cm, 2 cm wide at 16 cm, and 5 cm wide again at 20 cm depth. Such a situation is clearly intolerable.

The geometric aberration that causes the most difficulty is T_{122}. As a result of this coefficient, the beam becomes asymmetric about its centre, and the pion distribution becomes peaked to one side and suffers a decrease of intensity on the other side of the field. In general, it has been found that cross-coupling terms involving both X and Y planes such as T_{323}, T_{324}, and T_{144} are small and have no measurable effect on the beam. Two more terms that can have serious effects are T_{336} and T_{346}, both of which cause variations in Y beam size with energy. However, since the emittance of the beam in the Y plane (500 cm-mrad) is much larger than that in the X-plane (4.0 cm-mrad) the possible fractional change in beam size is much smaller for these aberrations than it is in the X-plane.

There are two basic methods for correcting these second order effects, both of which involve adding a sextupole component to the magnetic fields. It is possible to do this by curving the entrance and exit pole edges of the bending magnets or by actually adding physical sextupole magnets to the system. The latter has the advantage of being remotely adjustable once the system has been installed. Since the second order coefficients are linear functions of the sextupole fields,[4] minimization of the second order effects is essentially a linear programming problem once the second order coefficients and the effects of the sextupole fields are known, and this is the procedure that was used for this channel. It can be shown that there are several equally good solutions to the second order problem when four pole-edge curvatures and two sextupole field strengths are used as variables. The improvement in uniformity after second order correction can be seen by comparing Fig. 2 which shows the uncorrected distributions, with Fig. 3, which shows the pion distributions after second order correction. It may also be shown that the chromatic aberrations T_{126}, T_{336}, and T_{346} may not be eliminated simultaneously by any values for the sextupole fields, so some diaphragming of the pion beam adjacent to the irradiation location must be used to improve the lateral cut-off of dose.

<u>References</u>

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Table 1. System Parameters. The D's represent field-free drift regions, the Q's are quadrupole magnets, the S's are sextupoles, and the B's are uniform field bending magnets.

Element	Effective length (m)	Pole tip field (kG)	Aperture radius (cm)
D	0.971		
Q1	0.264	2.743	10.16
D	0.265		
B1	0.628	8.627	7.62 x 15.24
D	0.438		
S	0.254	<1	15.24
D	0.158		
Q2	0.400	3.716	15.24
D	0.336		
Q3	0.400	4.540	15.24
D	0.336		
Q4	0.400	3.716	15.24
D	0.158		
S	0.254	<1	15.24
D	0.438		
B2	0.628	8.627	7.62 x 15.24
D	0.300		
Q5	0.300	<5.00	10.16
D	1.360		

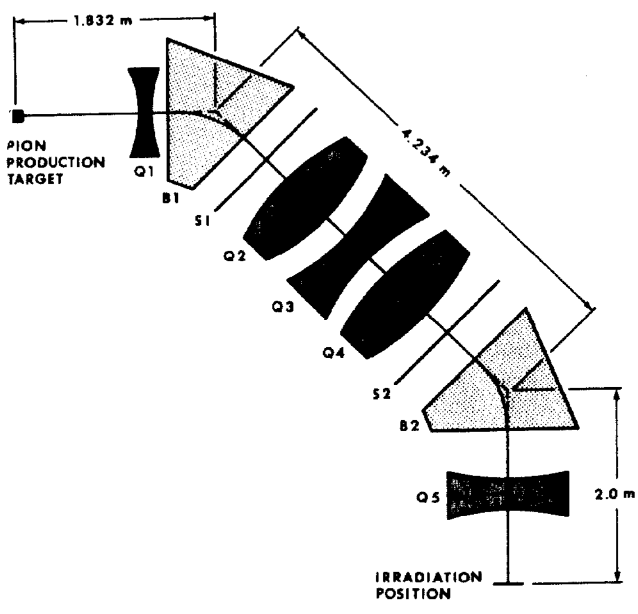

Fig. 1. Layout of the beam transport system in the plane of the two bends (equivalent light optics).

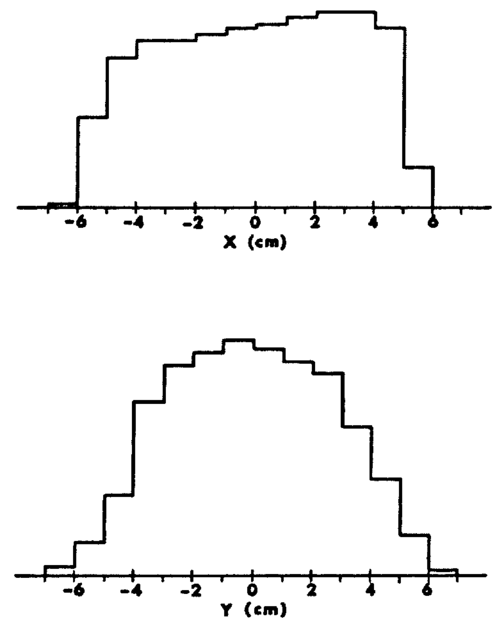

Fig. 2. Monte Carlo calculations of the pion beam distributions at the irradiation position before uniformity corrections.

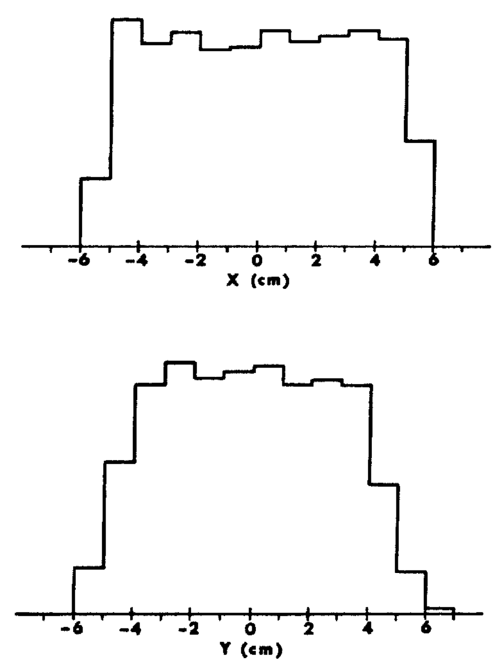

Fig. 3. Pion beam distributions after corrections up to second order are added.