CONSIDERATIONS FOR DESIGN PARAMETERS FOR A DEDICATED MEDICAL ACCELERATOR

Jose R. Alonso

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Jose R. Alonso, Ph.D.

Accelerator Division,
Lawrence Berkeley Laboratory
University of California
Berkeley, CA 94720

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Optimal Parameters of Relativistic Heavy Ions for Radiotherapy Research

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Introduction

There are only a very few critical parameters which determine the size, performance and cost of a heavy ion accelerator. These are the mass of the heaviest ion desired, the maximum range of this heaviest ion in tissue, and the highest intensity desired. Other parameters, such as beam emittance, beam delivery flexibility, reliability and experimental facility configurations are important, but are not primary driving factors in the design effort.

To evaluate optimal choices for these critical parameters, we shall examine the various clinical applications for a heavy ion accelerator, detailing the most desirable beams for each application.

Clinical Applications - Diagnostic

1. Radiography and Tomography

Heavy ion radiography has the demonstrated capability of differentiating much smaller soft tissue density variations than is possible with X-rays, offering greatly enhanced tumor detection possibilities. A beam particle passing through the body will lose energy determined by the density of material it traverses. Thus each particle, when it exits the body, has information corresponding to the electron density integrated along its path. In other words, as seen in Figure 1 the particle shown going through bone has less energy leaving the body, and will stop sooner in the detecting medium. The density resolution is limited to the accuracy to which one can measure the range of the exiting particles.

Beam Parameters. The first obvious requirement is that the beam must be able to pass through the thickest sample to be radiographed. Then, since image quality depends on the number of particles stopping in the detector, one desires low Z ions to maximize the particle count for a given patient dose. (Dose increases as the square of the nuclear charge, $Z^2$.) There is a limit as to how low in Z one should go, given by range straggling and multiple scattering, two effects which contribute to image quality degradation. Figure 2 illustrates these effects. Multiple scattering tends to broaden the effective track width, due to encounters
with electrons and nuclei in the material being traversed. The effect of this is to degrade lateral resolution of the image. Range straggling refers to the statistical spread in the stopping points of the particles. This effect degrades tissue density determination. Figure 3 shows how these two phenomena depend on the mass of the ion used. One sees that carbon appears to be a dividing line for both effects; anything lighter is clearly inferior, while gains are only relatively modest in going to heavier ions. The ideal choice of particle is best determined by the inherent resolution of the detector used. A heavier particle will add unnecessarily to the patient dose without great gains in detectable image quality.

A word is in order about beam delivery for radiography. Since there is an unavoidable separation between patient and detector, it is most important that the beam quality be as good as possible, i.e. that the beam have the smallest possible emittance. Flight paths between beam shaping apparatus and patient should be as long as possible, preferably in vacuum, with little or no scattering material in the beam. This will ensure the sharpest possible image.

Let us summarize, then, the requirements for radiographic and tomographic beam applications.

<table>
<thead>
<tr>
<th>Application</th>
<th>Ion</th>
<th>Range</th>
<th>Intensity</th>
<th>Beam Delivery</th>
</tr>
</thead>
<tbody>
<tr>
<td>Radiography</td>
<td>Carbon</td>
<td>40 cm min</td>
<td>low</td>
<td>Large field,</td>
</tr>
<tr>
<td></td>
<td>or heavier</td>
<td>(10^7)</td>
<td></td>
<td>good emittance</td>
</tr>
</tbody>
</table>

2. Radioactive Beams

Highly purified beams of positron emitters, such as ^11^C, ^15^O, ^19^Ne, have great diagnostic value. By pinpointing the stopping point in a patient the therapist can verify the treatment plan, and determine that the beam is going into the desired spot. Furthermore, these beams are the ultimate form of a "carrier free" tracer, capable of being instantly produced and implanted into any organ of the body in sufficient concentration to be easily followed with conventional diagnostic radioisotope detectors. Initial experimentation with these
beams at the LBL Bevalac has been highly successful, and refinements in beam delivery and experimental techniques are continuing.

**Beam Production.** These radioactive beams are produced in what is known as a peripheral nuclear fragmentation reaction\(^4\), where the projectile ion, \(^{12}\)C say, barely grazes by the target nucleus. The carbon nucleus is excited enough to emit one or more nucleons, but since the beam energy is so much higher than the energy transferred in the nuclear reaction, the projectile is hardly deflected from its original course. Production efficiency for \(^{11}\)C can be quite high, as is seen in Figure 4. For a beryllium target (lowest possible Z), almost 2% of the incident carbon beam can be converted to radioactive \(^{11}\)C. The curve falls off for thicker targets because the \(^{11}\)C itself starts interacting with target nuclei.

Emerging from the target, then, will be unreacted primary \(^{12}\)C ions, plus a host of nuclear reaction products, \(^{11}\)C being one of the prime constituents. All the emerging particles will have about the same energy-per-nucleon, and so constitute a mixed ion, roughly mono-energetic beam. A suitable magnetic analysis system is all that is required now to separate and purify the \(^{11}\)C beam. At the Bevalac we use two 16 degree bends with a pair of slits at an intermediate focus to separate the beam, and to provide a momentum-recombined focus at the desired point in the Treatment Room. Figure 5 shows a Bragg curve for \(^{11}\)C, verifying that essentially all \(^{12}\)C ions, which have a longer range than the \(^{11}\)C ions have been rejected.

A total efficiency of about 0.3% is typical, that is one radioactive nucleus delivered into the target volume for every 300 primary particles extracted from the Bevalac. This translates into about \(10^7\) \(^{11}\)C ions into a 1 cm\(^2\) area, a dose rate of about 10 rad per pulse. The specific activity for \(^{11}\)C approaches 0.2 microcuries per pulse; shorter half-life isotopes have even higher specific activities.

The efficiency at the Bevalac is as low as 0.3% because of constraints on target placement; we can capture only about 10% to 15% of the ions produced. Designing a dedicated radicisotope production line, where the target can be placed much closer to a quadrupole magnet, will
allow for capture over a much higher solid angle, potentially increasing the capture and transport efficiency to about 70%.

A summary of accelerated beam parameters for optimum radioactive beam production is given below.

<table>
<thead>
<tr>
<th>Application</th>
<th>Ion</th>
<th>Range</th>
<th>Intensity</th>
<th>Beam Delivery</th>
</tr>
</thead>
<tbody>
<tr>
<td>Radioactive beams</td>
<td>C, O</td>
<td>20 cm</td>
<td>high,</td>
<td>Good isotope separation capability.</td>
</tr>
<tr>
<td></td>
<td>Ne</td>
<td></td>
<td>$10^{10}$/sec</td>
<td></td>
</tr>
</tbody>
</table>

Clinical Applications - Radiotherapy

1. Small Field Treatments

These include ocular, pituitary and other treatments requiring highly intense irradiations of small, well localized areas. Beams for these treatments must be tightly focused, fairly uniform, and very stable. These requirements are not difficult to meet, particularly if the patient field is outlined with a collimator, and the beam spot is somewhat larger than the collimator opening. The intensity loss is of no consequence, tightly focused Bevalac beams can deliver dose rates over 100 kilorad/minute.

The beam range required for these treatments varies from a few cm for ocular work to about 15 cm for focal lesion studies.

For the moment let us postpone the choice of ion to be used for these treatments. We can then summarize as before, the optimal parameter set.

<table>
<thead>
<tr>
<th>Application</th>
<th>Ion</th>
<th>Range</th>
<th>Intensity</th>
<th>Beam Delivery</th>
</tr>
</thead>
<tbody>
<tr>
<td>Small field Therapy</td>
<td>--</td>
<td>15 cm</td>
<td>medium</td>
<td>Tight focusing, good beam stability and field uniformity</td>
</tr>
<tr>
<td></td>
<td></td>
<td></td>
<td>$10^8$/sec</td>
<td></td>
</tr>
</tbody>
</table>

2. Large Field Treatments

The primary requirements for large field therapy are a field size of 30 cm diameter, a depth of at least 25 cm and a dose uniformity of at least ± 3% over the entire treatment field. These specifications have
evolved from treatment experience at the LBL 184" cyclotron and the Bevalac, and represent conditions suitable for treating all but a very small fraction of our referred patients.

The problem of delivering a beam with the size and uniformity desired is most interesting, and is worthy of a digression at this point. Four techniques come to mind for producing such large beams,

a) Quadrupole defocusing,
b) Scattering-foil, occluding-ring system,
c) Wobbler,
d) Scanning systems.

Quadrupole defocusing does produce large fields, but uniformity is not controllable, and is in general inadequate for radiotherapy. Beam emittance is preserved, making this mode of beam delivery acceptable for most radiographic applications at the Bevalac, where field non-uniformities of ±20% are acceptable.

The scattering-foil, occluding-ring system, pioneered at Harvard, is shown in Figure 6. The beam is first broadened with a lead scattering foil, then portions of the intense central region are masked with the rings, and a second scatterer at the rings diffuses the beam to fill in the holes produced. By proper selection of foil thicknesses and ring sizes a flat beam is obtained at the patient. This clever system has been used very successfully with protons at Harvard and alpha particles at LBL, but has proven to have certain limitations for Bevalac beams. The main reason is that the higher-energy heavier beams need much more lead to give the same scattering angles leading to large amounts of energy loss in the scatterers, and substantial beam losses due to nuclear fragmentation. Our 18 cm neon beam needs over 1 cm of lead, and requires 20% higher extraction energy to preserve the desired range in tissue. However for smaller fields, or with longer beam-preparation space this flattening technique is quick, reliable and almost foolproof.

The wobbler concept, shown in Figure 7 uses a rotating magnetic field to sweep the beam in a circular pattern. The rotating field can be implemented with a physically rotating magnet, or, more practically, by correctly energizing the coils of a multipole magnet. The figure
illustrates a sextupole magnet powered with 3 phase AC. The desired dose uniformity in the therapy field is obtained by setting the ratio of magnet strength (radius of swept circle) and the scattering foil thickness (beam size). Adequate field flatness is obtained with a continuously running wobbler, but somewhat higher beam utilization efficiency is obtained by sweeping a smaller diameter beam, then turning the wobbler off to fill in the central hole. The wobbling concept is used extensively with low energy beams, but has not yet been done with relativistic heavy ions. The scheme is practical, though, and could be implemented easily.

Beam scanning with independent horizontal and vertical magnets is shown schematically in Figure 8. This technique is a most versatile delivery system, and holds the greatest promise for developing the full potential of heavy ions for therapy. Scanning systems can range from very simple to very complex, the simplest being easy applications of known technology, while the most complex require substantial research and development in control and instrumentation, and extensive radiobiological work to prove their viability.

A simple scanning system is illustrated in Figure 9. The x magnet (fast sweep) runs in a free-running sawtooth pattern, sweeping a suitably broadened beam across the horizontal plane, while the slow magnet sweeps once through the field per beam pulse. Required to deliver the prescribed dose are a trigger at the start of each beam spill to initiate the slow sweep, and a controlled uniform spill. No feedback on the instantaneous beam position is needed, only the assurance that if the spill is flat the dose delivered will be uniform across the field. We are in the process of installing such a system at LBL, the magnets to be used are shown in Figure 10.

In the ultimate scanning system, the magnets would direct a pencil beam into a 3-dimensional volume element, dwelling there until the prescribed dose for that pixel is delivered, then moving on to the next pixel. The concept is elegant, and one can visualize coupling treatment planning computers with treatment delivery computers in a totally automated therapy system. However, a look at some numbers is essential.
to appreciate both the virtues and difficulties of this scheme. Minimum pixel size, determined by range straggling and multiple scattering, is of the order of 3mm x 3mm x 4mm. A typical treatment volume will contain around $10^5$ to $10^6$ pixels, so to keep treatment time to 2 minutes or less, dwell time in each pixel is 1 millisecond or less. Control, dosimetry and magnet response will all be pushed to the limits of present technology, and also questions must be addressed of patient safety in case of malfunction and high dose-rate effects (200 rads/millisecond = $10^7$ rads/min). Much study and development must be done before such a sophisticated scanner is ready for patients.

Returning now to the subject of beam parameters for the accelerator, let us address the point of the most desirable particle for therapy. This question has been one of the primary research goals of the Bevalac Biomedical group for several years now. Results of extensive experimentation indicate that there is no single optimal particle, the best ion to be used depends on the type, depth and character of the tumor being treated. Argon appears to be too heavy for deep treatments, (too much normal tissue damage), but OER's for light ions (C, Ne) are higher than one would like. For our purposes, specifying the longest range of the heaviest particle, let us take 25 cm silicon as our most demanding condition. We will see later that in the same accelerator substantially longer ranges are available for lighter ions, and heavier ions can be used, with shorter ranges.

Summarized below, then, are the critical parameters for large field therapy.

<table>
<thead>
<tr>
<th>Application</th>
<th>Ion</th>
<th>Range</th>
<th>Intensity</th>
<th>Beam Delivery</th>
</tr>
</thead>
<tbody>
<tr>
<td>Large field</td>
<td>Silicon</td>
<td>25 cm</td>
<td>high</td>
<td>30 cm field, $10^{10}$/sec, +3% uniformity</td>
</tr>
</tbody>
</table>

Summary of Specifications

In Table I below are collected all the specifications pertaining to accelerator design for the various applications discussed above. The Beam Delivery column is omitted since delivery techniques are not critical determinants of accelerator design parameters. Minor
adjustments in beam energy or spill control are driven by beam delivery methods, but these we will treat as engineering details.

<table>
<thead>
<tr>
<th>Application</th>
<th>Ion</th>
<th>Range</th>
<th>Intensity</th>
</tr>
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<tbody>
<tr>
<td>Radiography</td>
<td>Carbon</td>
<td>40 cm</td>
<td>low (10^7)</td>
</tr>
<tr>
<td>Radioactive beams</td>
<td>Neon</td>
<td>20 cm</td>
<td>high (10^{10})</td>
</tr>
<tr>
<td>Small field</td>
<td>Silicon</td>
<td>15 cm</td>
<td>medium (10^8)</td>
</tr>
<tr>
<td>Therapy</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Large field</td>
<td>Silicon</td>
<td>25 cm</td>
<td>high (10^{10})</td>
</tr>
<tr>
<td>Therapy</td>
<td></td>
<td></td>
<td></td>
</tr>
</tbody>
</table>

We will see that the large field therapy requirements place the greatest demand on the accelerator.

If one assumes an accelerator built to deliver 25 cm range silicon, Figure 11 shows the range of other ions accelerated in the same machine. One sees clearly that 40 cm carbon for example is easily obtained.

**Effects of these Parameters on Accelerator Design**

First of all, the injector type is determined by the requirement for high intensity which is most easily met by a linear accelerator. Cyclotrons are excellent for modest intensity long duty cycle beams, but do not look attractive for producing the high peak currents required for synchrotron injection.

The question of maximum range is not terribly critical; if more energy is wanted, the ring is made a bit bigger, and a few more magnets are added at only a modest cost increase. Figure 12, taken from the LBL/Arizona Medical Ion Accelerator Design Study of 1977\(^9\) illustrates this point. Although the y axis is 1977 dollars, the point is still clear that adding to the beam energy has little budgetary impact to the project.

What is a very important factor, though, is the particle mass, since this determines the required injector size. This comes about because it is most desirable to inject fully stripped ions into the synchrotron (minimizes the vacuum system specifications), and heavier ions require
greater energy to strip off the last, most tightly bound electrons. Figure 13 shows the injector energy needed to produce beams in which at least 25% of the ions can be stripped of all their electrons. One sees that a very modest 1 MeV/amu injector will suffice for carbon, whereas argon requires about 8 MeV/amu, a linac the size of the SuperHILAC. Since the injector can approach 50% of the total facility cost, the decision of which particle to be used should be carefully weighed.

If funding is anticipated to be a problem, perhaps the best approach would be to design an injector system for the heaviest ion anticipated, say a maximum energy around 10 MeV/amu, but which is divided into different stages. Building and installing the first stage only might produce an injection energy adequate for an intermediate ion, such as neon, then at a future date as funds and needs dictated other stages could be installed in their designed locations to upgrade the facility.
References


Figure Captions

1. Heavy ion radiography. A beam particle passing through a (compensated) sample loses energy according to the integral of the electron density along its path. The energy of the emerging particle is detected, either by its range in a plastic stack or by a signal from a particle detector. The energy variations throughout the sample are used to construct the radiograph.

2. Multiple scattering is seen as the lateral broadening of the envelope of a particle beam as it penetrates through a sample, caused by multiple collisions between beam particles and electrons in the medium. Range straggling refers to the spread in stopping points of a mono-energetic beam in a sample, caused by the statistical nature of the energy-loss process.

3. Range straggling and multiple scattering for different ion species. One sees the definite advantage for using carbon or heavier ions in radiography.

4. $^{11}$C production as a function of beryllium target thickness (energy loss of the primary beam). Beryllium offers the highest $^{11}$C yield per beam energy loss of any other solid target, and the lowest beam spreading due to multiple scattering. The production curve falls off for thicker targets because of $^{11}$C loss through further nuclear reactions, coupled with depletion of the primary beam.

5. Bragg curve for purified $^{11}$C beam. Notice the total absence of any $^{12}$C peak beyond the $^{11}$C peak. From these data, and the known production efficiency for this run, one estimates less than 1 part in $10^5$ contamination from the primary beam.

6. Occluding ring beam delivery system. Beam is scattered, portions blocked and further scattered to produce large uniform fields for radiotherapy.

7. Beam wobbler system. A sextupole magnet is powered with 3 phase AC to produce a rotating bending field, sweeping the beam in a circle. Field flatness is obtained by adjusting the ratio of the sweep radius and the beam size. Best beam utilization is obtained by filling in the central hole with the wobbler off.
8. Beam scanning system. Independent x- and y-axis magnets move the beam across the treatment field. Advanced scanning concepts incorporate the range modulator and fast beam on/off switch into the overall control system.

9. Free-running scanning system. A large-diameter beam spot is swept across the treatment field in the pattern shown. Time structures for the scanning magnetic fields and beam intensity are shown below. The flat spill is obtained by fast-feedback control of the accelerator beam extraction system.

10. Magnets to be installed for the Bevalac scanning system. The fast magnet (smallest of the two) is capable of a peak-to-peak swing in 1.2 milliseconds.

11. Range of different ions accelerated in the same machine, showing significant range gains for lighter ions.

12. Hardware costs for circular accelerators, taken from the LBL Dedicated Medical Ion Accelerator Design Study (Ref. 8, page 65a). Curves A through D represent various synchrotron options, the rapidly ascending curves to cyclotron designs. The main point to note is that synchrotron costs do not increase very rapidly as the desired energy is raised.

13. Beam energy (related to injector size) required to leave 25% of the ions with no electrons after passing through a stripping foil.
Figure 1
Multiple Scattering

Range Straggling

Figure 2
Figure 3

Range Straggling

\[ \sigma = \sqrt{\langle R^2 \rangle - \langle R \rangle^2} \] (g/cm\(^2\))

\[ \text{Mean range (g/cm}\,^2\)\]

Multiple Scattering

\[ \text{Mean beam deflection (g/cm}\,^2\)\]

\[ \text{Mean range (g/cm}\,^2\)\]
Production efficiency and multiple scattering (Beryllium target)

$^{12}\text{C} \rightarrow ^{11}\text{C}$

Figure 4
Figure 5

XBL 793–753
1st scatterer

2nd scatterer

Rings

Tumor

Beam delivery Occluding ring system

Figure 6
Sextupole wobbler magnet

Pb foil

±1% flat region contains 40% of beam

Beam delivery wobbler system

Figure 7
Figure 9

Beam spot size

Treatment field

X magnetic field

Y magnetic field

Beam pulse

Time 0 to 1 second
Beam Rigidity

90 kG-m

Figure 11
Figure 12

HEAVY ION RANGE - CM.

<table>
<thead>
<tr>
<th>Element</th>
<th>Range (cm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Helium</td>
<td>137.5</td>
</tr>
<tr>
<td>Carbon</td>
<td>45.8</td>
</tr>
<tr>
<td>Nitrogen</td>
<td>39.3</td>
</tr>
<tr>
<td>Oxygen</td>
<td>34.5</td>
</tr>
<tr>
<td>Neon</td>
<td>27.5</td>
</tr>
</tbody>
</table>

**ISOC =Isochronous**

**SC = Superconducting**

**FM = Frequency Modulated**

**ε = Charge to Mass Ratio**

**Hardware Costs in Millions of Dollars**

**B** = IN TESLA METERS

**T(ε=1.0)** MEV/AMU

**T(ε=0.5)** MEV/AMU
25% of ions fully stripped

Figure 13