COMPACT CYCLOTRON ENGINEERING FOR APPLICATION

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ABSTRACT

Some current uses of compact cyclotrons in the fields of physics, nuclear medicine, and neutron therapy are considered. The principle cyclotron performance characteristics required by each of these applications are discussed and design innovations introduced to meet these requirements presented.

INTRODUCTION

Dave Judd aptly pointed out in Gatlinburg that the cyclotron is viewed differently by different groups. One viewpoint which Dave did not touch on, however, was the "cyclotron as seen by the cyclotron manufacturer." Today I will endeavor to fill that gap.

The principal problem faced by a manufacturer of compact cyclotrons is to produce a simple economical machine which can be used in a wide variety of applications. These applications which range from physics and radioisotope production to neutron therapy make widely varying demands on the cyclotron. Thus the radioisotope producers usually want as much energy and external beam current as possible while the physicists want variable energy negative ion acceleration and extremely good emittance. The therapists, and everyone else for that matter, want high reliability. To develop a basic compact cyclotron which can be economically adapted to these varying requirements has indeed been an interesting task.

Over the last seven years we have built thirteen compact cyclotrons. Of these, seven are used for radioisotope production, five for physics, and one for neutron therapy. My intention is to review these applications with you and to share with you our approach to some of the engineering problems they have presented. Let us start by first looking at one of the current basic compact cyclotron designs.

A BASIC COMPACT CYCLOTRON

The CS-30 is our most current basic compact cyclotron. To date two of these machines have been built; one for radioisotope production and the other for neutron therapy. The basic specifications of this machine are given in Table I. A more detailed description of the components which make up our compact cyclotrons can be found in the literature.1,2,3,4
Table I Basic Specifications of CS-30 Cyclotron

<table>
<thead>
<tr>
<th>Specification</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Type of Cyclotron</td>
<td>Isochronous</td>
</tr>
<tr>
<td>Proton Energy</td>
<td>26.2 MeV</td>
</tr>
<tr>
<td>Deuteron Energy</td>
<td>15.0 MeV</td>
</tr>
<tr>
<td>Helium Three Energy</td>
<td>39.0 MeV</td>
</tr>
<tr>
<td>External Beam Power</td>
<td>2000 Watts</td>
</tr>
<tr>
<td>Pole Diameter</td>
<td>38 Inches</td>
</tr>
<tr>
<td>Number of Sectors</td>
<td>3</td>
</tr>
<tr>
<td>Weight</td>
<td>22 Metric Tons</td>
</tr>
<tr>
<td>Number of Dees</td>
<td>n</td>
</tr>
<tr>
<td>Acceleration Mode</td>
<td>Fundamental</td>
</tr>
<tr>
<td>Maximum Voltage Gain per Turn</td>
<td>$10^5$ Volts</td>
</tr>
<tr>
<td>Price</td>
<td>$390,000</td>
</tr>
</tbody>
</table>

The appellation CS-30 denotes a fixed energy cyclotron having an external beam with a maximum mass-energy product of 30.

The basic CS-30 has been modified by the addition of an external negative ion source, an axial injection system, and a negative ion extraction system to form a negative ion cyclotron which we call CNI-30. This negative ion machine was recently completed and is now serving as an injector to an EN Tandem Van de Graaff at the Australian National University.

The basic CS-30 has also been modified by the addition of profile coils and a remotely adjustable RF system to form a variable energy cyclotron which we designate as CV-28. The maximum mass energy product of the CV-28 is less than that of the CS-30 due to the necessity of removing pole iron in order to install the profile coils. Two CV-28 cyclotrons are currently under construction.

The concept of developing a basic cyclotron design and reusing components and concepts as often as possible has played a key role in the economical development of compact cyclotrons. Let us now look at some compact cyclotron applications and the design problems they presented.

**RADIOISOTOPE PRODUCTION**

Within the last few years we have installed five compact cyclotrons which are now routinely producing radioisotopes for use in medicine. Before the end of the year we will complete two additional medical cyclotron installations. The isotopes which can be produced by these cyclotrons are currently used in a wide range of clinical applications. Generally these clinical applications fall into four groups which Glass and Silvester have summarized as follows.

**Localization Studies.** These studies include use of $^{18}$F to detect metastasis in bone, $^{81}$Rb to localize and detect enlargement of the spleen, $^{123}$I to detect thyroid nodules in children, and $^{67}$Ga for detection of soft tissue tumors.
Estimation of Body Spaces. These studies include use of $^{11}$C to estimate red cell volume and $^{43}$K to estimate the exchangeable potassium in multiple electrolyte studies.

Measurement of Blood Flow. These measurements include use of $^{82}$Rb to determine myocardial blood flow and $^{15}$O to determine cerebral blood flow and metabolism.

Study of Organ Function. These studies include use of $^{81}$Rb to estimate rate of red cell destruction by the spleen, $^{123}$I for the thyroid function in children and in pregnancy and for cardiopulmonary function, $^{15}$O for lung function and organ blood flow and oxygen, $^{11}$C for assessment of cardiac shunts and lung function, and $^{13}$N for lung function studies.

The amounts of radioisotope administered depend on a number of factors including the half life of the isotope and the rate at which it is eliminated from the body. The amounts generally administered vary between a few hundred microcuries and 5 milli-curies.

The reactions used to produce several commonly used radioisotopes as well as the production rates achieved are summarized below. This data has been compiled in cooperation with several of the groups as indicated by the references.

Fluorine $^{18} - 110$ min. Sloan-Kettering Institute$^6$ reports a yield rate of 6 mCi/$\mu$A-hour using 20 MeV $^3$He$^+$ ions on distilled water. Typical runs yield 70 mCi/30 min at 25 $\mu$A in a 10 ml volume of water.

Argonne Cancer Research Hospital$^7$ reports a gross yield of 10 mCi/$\mu$A-hour using 8 MeV deuterons on a Ne$_2$0 gas target. About 85% can be washed out of the quartz-lined target chamber. ARCH has not attempted to operate at high currents or produce more than a few mCi due to personnel hazard problems. Hammersmith Hospital in London plans to switch to this technique, possibly using a continuous flow system. The yield would be limited by window heating. Presently 70 $\mu$A to 80 $\mu$A is a practical range.

Gallium $^{67} - 78$ hours. ORNL$^8$ reports a yield rate of 600 mCi/$\mu$A-hour using 8 MeV deuterons on Zn$^{66}$. No data is given on the gross yield, but an internal target should withstand 30 - 100 $\mu$A to provide 1 to 3 mCi per hour. Sloan-Kettering reports a similar yield rate, and believes that 100 $\mu$A is a practical internal target current.$^9$

Indium $^{111} - 2.8$ days. ORNL$^{10}$ reports a yield rate of 600 mCi/$\mu$A hour. Gross yield should be 30 mCi/hour using a 50 $\mu$A beam of 15 MeV protons on Cadmium 111 enriched to 95-98% abundance. The ORNL 86-inch cyclotron was used to deliver a 200 $\mu$A beam of 22 MeV protons to an internal water cooled powder target. Net energy at the target was 15 MeV after passing through the water cooling tube, and target geometry was such that 50% of the measured current struck the target material.

Sloan Kettering Institute reports$^9$ a yield rate of 420 mCi/$\mu$A hour, and believes an external target could be designed for 100 $\mu$A operation to yield 42 mCi/hour.
Iodine 123 - 13.3 hours. Sloan-Kettering Institute\textsuperscript{11} has studied various reactions, and concludes that for their CS-15 cyclotron the most efficient method is to use 15 MeV protons on 77% enriched Te\textsuperscript{123}. Their measured yield rate is 5 mCi/\mu A hour, with a gross rate of 30 mCi/hour to produce 100 to 200 mCi batches.

Brookhaven National Laboratory\textsuperscript{12} reports a yield of 726 mCi/\mu A hour using 30 MeV He-3 ions (CS-22 cyclotron energy) on 77% enriched Te\textsuperscript{123}, in a continuous flow target system. This reaction produces Xe\textsuperscript{123} which is purified to remove unwanted fractions of the iodine isotopes before decaying to I-123.

Carbon 11 - 20.4 min. Sloan-Kettering Institute\textsuperscript{11} generates Carbon-11 with a variety of gas targets to produce \( ^{11} \text{CO}_2 \) and \( ^{11} \text{CO} \) with yields ranging from 100 mCi/liter to 3000 mCi/liter, beam intensities up to 15 \mu A. Flow rate is nominally one liter per minute at N.T.D.

Iron-52 - 8 hours. Sloan-Kettering Institute\textsuperscript{11} reports a yield of 0.7 mCi/\mu A hour using He-3 on chromium. Saturation yield is calculated to be 8.4 mCi/\mu A hour. At 100 \mu A on an internal target, a practical yield would be 70 mCi/hour.

Rubidium 81 - 4.7 hours. Sloan-Kettering Institute\textsuperscript{11} reports a yield rate of 35 mCi/\mu A hour for He-3 on a Na Br target. At a current of 25 \mu A on a covered external target, a practical yield would be 835 mCi/hour.

Technetium 99m - 6.0 hours. Beaver and Hupt\textsuperscript{13} project a yield rate of 32 mCi/\mu A hour of Tc\textsuperscript{99m} and 600 mCi/\mu A hour of Mo\textsuperscript{99} using 22 MeV protons from a CS-22 cyclotron, with internal targets. They believe that currents up to 500 \mu A can be used, for a gross yield of 15 Ci/hour of Tc-99m.

For the CS-15, yield rate of Tc-99m would be 11 mCi/\mu A hour, and of Mo-99, 100 mCi/\mu A hour.

ATTAINING HIGH BEAM POWER FOR RADIOISOTOPE PRODUCTION

The primary requirement for an isotope production cyclotron is a routinely available external beam of 50 to 100 microamperes. The New England Nuclear cyclotron, for example, routinely runs an external 22 MeV proton beam of more than 90 microamperes, and has run external proton currents as high as 160 \mu A.

Our principle problem with the extraction of high beam currents is the high beam density at the extractor due to the strong vertical focussing in the fringe field where the extractor is located. To overcome this difficulty we first followed the customary procedure of making a long tapered notch in the septum in an effort to spread the beam out horizontally. As the notch became long enough to be effective we had serious degradation of the radial beam quality due to the dependence of deflection on beam height. Further, the tip of the notch always had a tendency to fall. With this type of septum we found ourselves limited to a reliable external beam of somewhat less than one kilowatt. In order to overcome this difficulty we developed what we call a "preseptum." This is
a curved sloped tungsten knife edge which shadows the septum from the incoming beam. The preseptum has no electrostatic field behind it so that we may use a very gradual slope without destroying the radial beam quality. The main septum now requires no notch at all which improves radial beam quality. The preseptum is a simple assembly which is fastened by a single clamp and can be removed and replaced with a manipulator. With the addition of the preseptum we found we could reliably extract about 1700 watts of external beam. Since the preseptum which is easy to replace fails rather than the main septum, radiation doses received during extraction system servicing are reduced.

To further increase the external beam which can be reliably extracted and to reduce the power density on our internal target we have developed another device which we call the vertical beam expander. This is simply a pair of horizontal plates, one above the beam and one below the beam. An RF potential having the same frequency as the axial betatron oscillation near the extraction radius is applied between them. Typically enough voltage is applied to reduce the beam density by a factor of two. Use of this device allows reliable external beam power well in excess of 2000 watts.

One of the consequences of the precessional extraction system is that the beam density is a function of radius. Operationally this causes a problem because a slight adjustment of the controls may place a dense portion of the beam at the septum which results in septum failure if the machine is operating at high power levels. To overcome this difficulty we developed a light monitoring system which monitors the light emitted by the cyclotron's tungsten pre-septum. This device has proved a valuable tuning aid and may be interlocked to prevent septum meltdown.

Using the preseptum, the vertical beam expander and the light monitoring system we have achieved external beam powers as high as 3600 watts with reliable routine operation of about 2500 watts. The ability to reliably extract large external beam currents has been a key economic factor in the three cyclotrons we have built for commercial isotope producers.

PHYSICS APPLICATION

Five of the cyclotrons we have built are being used for physics applications. Three of these are negative ion cyclotrons which are used as injectors to tandem electrostatic accelerators. The remaining two are variable energy cyclotrons which are presently under construction. The variable energy machines will be used for neutron physics and nuclear structure work.

The negative ion cyclotrons are high energy sources of negative ions for tandem electrostatic accelerators. Due to the relatively high gas flow required for the production of negative ions and the high stripping cross-section of negative hydrogen ions the cyclotron source is located outside the acceleration chamber and the negative ions axially injected. In this way the pressure
in the cyclotron may be maintained below $10^{-5}$ Torr. Our experience indicates that beam loss due to stripping becomes appreciable in the region of 1 to $2 \times 10^{-5}$ Torr. In addition to the vacuum considerations, the design of the positive extraction electrode required for negative ion extraction is rather critical in that care must be taken to prevent Penning discharges and other forms of electron loading.

In May we completed installation and testing of our largest negative ion cyclotron at the Australian National University. This cyclotron has an $H^-$ energy of 26.2 MeV and a $D^-$ energy in excess of 14 MeV. Its energy resolution without phase limitation is less than 50 KeV FWHM and its emittance is less than 20 mm-mr in the horizontal and vertical plane. Maximum external $H^-$ current is of the order of 20 μA external.

We are also building variable energy cyclotrons for physics applications. The variable energy cyclotron we now have under construction for Physikalisch-Technische Bundesanstalt, Braunschweig, West Germany is designed to operate in the single turn extraction mode with pulse selection in the cyclotron center. Due to the high magnetic field and relatively low dee voltage, phase selection by means of a slit system eliminates a great deal of the beam lying within the correct phase interval. For this reason we are employing a radio frequency phase selector which deflects the beam upward or downward depending on whether it is leading or lagging the desired phase. The phase is then limited by introducing horizontal collimator slits. Since the aperture of these slits may be greater than the inherent beam height the center of the phase interval is transmitted unattenuated by the phase selection system. Degradation of the axial beam quality is reduced by making the axial deflection field include approximately $1/\sqrt{2}$ turns.

It is noteworthy that this pulsed cyclotron will be mounted on a carriage which moves on a tract concentric with a neutron producing target. A long neutron time of flight spectrometer remains stationary and the cyclotron moved in order to make differential cross-section measurements. We hope to achieve as much as $3 \times 10^{-12}$ coulombs per pulse from the cyclotron to facilitate these measurements.

GOOD GEOMETRICAL BEAM QUALITY

Good geometric beam quality which is desirable to reduce beam transport system costs becomes essential when the beam is to be directed through a tandem electrostatic accelerator. For this reason we have worked hard to improve the geometric beam quality of the external beam. As the cyclotron beam passes through the electrostatic deflector and subsequently through the cyclotron fringe field it is radially defocussed as well as dispersed. Since cyclotron fringe fields tend to be anything but linear, sextupole and higher order field components operate on the beam. The wider the beam becomes the more strongly these components are coupled. Unless one
corrects for these nonlinear effects outside the cyclotron the effective emittance of the beam is increased.

Our approach has been to minimize nonlinear effects by keeping the exiting beam narrow in the radial direction, thus reducing the coupling of nonlinear field components. This is accomplished by a series of magnetic channels which locally reverse the gradient of the cyclotron fringe field and produce radial focusing. In the Australian National University cyclotron, for example, we use two magnetic channels; one located on the electrostatic deflector and one about halfway between the electrostatic deflector and the exit port. Careful design of these channels allows us to get 90% of the beam in a full area phase ellipse with an area less than 15 mm-mr.

We have also found it useful to make the linear field area of one of the magnetic channel sections substantially wider than the beam. We can remotely vary the radial position of this section which allows us to steer the beam at the cyclotron exit port.

**Neutron Therapy**

One of our cyclotrons which is currently completing tests in Berkeley will be used primarily for neutron therapy. This machine which will be installed early next year at Tokyo University, will be used in a neutron therapy program similar to that which has been underway at Hammersmith Hospital in London for the last several years.15

The 15 MeV deuteron beam from this machine will produce fast neutrons via the $^{9}$Be($d$,n)$^{10}$B reaction. Even though the deuteron energy is about 1 MeV less than that used at Hammersmith we expect a depth for 50% dose of 8.5 cm for a 10 cm x 10 cm field with a source to skin distance (SSD) of 125 cm, using a thick beryllium target. Table II which was prepared by Dr. John Parnell16 of Hammersmith summarizes dose rates and depths for 1/2 dose for various field sizes, target thicknesses, and source to skin distances.

Table II  Neutron Therapy Data for CS-30 Cyclotron

<table>
<thead>
<tr>
<th>Reaction: $^{9}$Be($d$,n)$^{10}$B</th>
<th>Deuteron Energy: 15.0 MeV</th>
</tr>
</thead>
</table>

**Thick Target**

<table>
<thead>
<tr>
<th>SSD (cm)</th>
<th>Dose Rate (Rad/min/100 μA)</th>
<th>Depth for 50% of Dose at 0.5 cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>5 x 5</td>
<td>10 x 10</td>
<td>20 x 20</td>
</tr>
<tr>
<td>field</td>
<td>2 x 5</td>
<td>10 x 10</td>
</tr>
<tr>
<td>125</td>
<td>35 ± 3</td>
<td>40 ± 3</td>
</tr>
<tr>
<td>160</td>
<td>22 ± 2</td>
<td>24 ± 2</td>
</tr>
<tr>
<td>field</td>
<td>23 ± 2</td>
<td>24 ± 2</td>
</tr>
<tr>
<td>125</td>
<td>7.4 ± 0.3</td>
<td>8.9 ± 0.3</td>
</tr>
<tr>
<td>160</td>
<td>7.7 ± 0.3</td>
<td>9.2 ± 0.3</td>
</tr>
<tr>
<td>field</td>
<td>10.9 ± 0.3</td>
<td>10.5 ± 0.3</td>
</tr>
</tbody>
</table>
Thin Target (70 mg/cm²) Ta backing

<table>
<thead>
<tr>
<th>SSD</th>
<th>Dose Rate (Rad/min/100μA)</th>
<th>Depth for 50% of Dose at 0.5 cm</th>
</tr>
</thead>
<tbody>
<tr>
<td>125 cm</td>
<td>18 ± 3</td>
<td>8.3 ± 0.5 9.8 ± 0.5 11.1 ± 0.5</td>
</tr>
<tr>
<td>160 cm</td>
<td>11 ± 3</td>
<td>8.0 ± 0.5 10.1 ± 0.5 11.4 ± 0.5</td>
</tr>
</tbody>
</table>

The work of Goodman and Brennan, and Bewley and Parnell have been particularly helpful in estimating the value of compact cyclotrons to current neutron therapy programs. It must be observed that while dose rates of 20 rads per minute are generally believed to be adequate for neutron therapy purposes, depth for one half dose distances of less than 10 cm are not. D-T fusion generators which produce 14 MeV neutrons can provide deeper depths for one half dose greater than 10 cm; they have not yet however produced steady state neutron source strengths of 8 x 10^12 n/sec 4% which is required to provide 20 rad per minute in 10 cm x 10 cm field 125 cm from the source. Until such a generator is built neutron therapy programs must rely on cyclotrons to produce fast neutrons. Compact cyclotrons will have a role in neutron therapy programs where the tumor to be treated is not deeply seated.

In order to increase the usefulness of the cyclotron as a therapy tool we are developing an "isocentric" neutron therapy system. This device which is similar to isocentric devices used with electron linacs allows moving the collimated neutron beam concentric with a tumor without moving the patient. Development of this type of equipment as well as target systems and dosimetry systems is typical of the challenges facing the cyclotron manufacturer today.

CONCLUSION

We have seen that compact cyclotrons have proven useful for physics and radioisotope production and may have an application in neutron therapy programs. The design and performance of these compact machines has made significant advances in order to satisfy these applications. What the future may hold is hard to say. On the one hand the isotope producers and therapists are demanding machines with increased energy and beam power while the physicists are pressing for better energy resolution and emittance. If the short lived gasses 11C, 13N, 15O come into general use there may be a demand for an ultra compact self shielded cyclotron for general use in hospitals.

Whatever the outcome of these trends it is the job of the cyclotron manufacturer to be ready with equipment which can economically meet these requirements.
REFERENCES


7. Private Communications, P.V. Harper, Argonne Cancer Research Hospital, Chicago, Illinois


10. Private Communication, J. Beaver and H. Hupf, ORNL, now at Mt. Sinai Hospital, Miami, Florida


13. "Production of Tc-99m on a Medical Cyclotron: A Feasibility Study", J. E. Beaver and H. B. Hupf, Division of Nuclear Medicine, University of Miami, School of Medicine, Mt. Sinai Medical Center, Miami Beach, Florida. Submitted to J. Nuclear Medicine


DISCUSSION

WEGNER: Could you comment just a little bit more on the outstanding problems of making a hospital cyclotron with an on/off switch?

HENDRY: I think I can answer that in almost one word—they are formidable! The first thing that we would do is limit the cyclotron to a single particle, single energy machine. At this time it looks like it would be deuterons, somewhere around 6 to 8 MeV. My feeling is that as far as the cyclotron is concerned that simplifies things a great deal. It would have an external beam. We all know deuterons are quite easy to extract. The major area that concerns me in the simple box with an on/off switch approach is actually what do the isotope-producing targets look like and how best do we interface the cyclotron with some rather complicated chemical systems.

WILLAX: Have you made a test of your pulsing device, and if so could you explain a little more about it?

HENDRY: I thoroughly expected that question. No, we have not yet made a test of the pulsing device. It is scheduled to be tested in January of next year. The components are currently being built. We have made some fairly detailed computer studies as well as analytical studies of how the beam behaves in the electric field set up by the pulse shortening RF field. The results of these studies indicate that it is practical to get a pulse with a width of about 0.5 nsec at the base line. This is substantially better than we need. This would occur for a potential of about 40 kV across a 1 cm plate spacing. It is also interesting to note that the deflection is related to the cube of the length of the plates.

FINLAN: Can you tell me if you considered building a four-sector, separated sector, cyclotron for large external beam currents?

HENDRY: I don't feel that the question of three or four sectors is an extremely important factor in the amount of external beam current we can get. We initially chose a three-sector design.

FINLAN: Sorry, I should have said 'separated' sector only.
HENDRY: No, we haven't really considered getting into that at this point. We are closely watching what is happening in Bloomington and SIN. Some of the problems are quite beyond the scope of the manufacturer. We must wait for the academic community to solve most of the major problems before we get involved, because I just don't think we have the resources to build that class of machine from scratch.

MORRISON: Fluorine-18 delivered in Vancouver costs us about $60 a patient for bone scanning, and I suspect that perhaps your choice of fluorine might have been better in that technetium may now be used as a phosphate, phosphonate or a polyphosphate for bone scanning with physical characteristics which are considerably better detection-wise than is fluorine. I wonder what your comments are regarding the use of fluorine vs technetium polyphosphate for bone scanning, because using technetium, bone scanning is now available to virtually every hospital in the world that can afford a molybdenum-technetium generator.

HENDRY: I think I have to answer the question on two levels. Wearing the hat of a cyclotron manufacturer, of course, we would like to see as many applications as possible for compact cyclotrons. On the other hand, as far as the art of nuclear medicine goes, I think we all as human beings are interested in seeing it progress in the best way that it can.