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THE NARROW PROTON BEAM THERAPY UNIT AT THE THE SVEDBERG LABORATORY IN UPPSALA

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Abstract

The synchrocyclotron at the The Svedberg Laboratory (TSL) in Uppsala is now reconstructed and can presently operate with fixed frequency and proton energies up to 100 MeV. A first treatment room with a narrow proton beam unit for therapy of eye tumours is now in operation. Therapy of eye melanomas started in April, 1989 and during 1989 and 1990, 19 patients were treated with 72 MeV protons. The narrow beam unit provides a fixed horizontal beam and the patient is treated in a seated position. The present paper describes mainly the technical aspects of the unit which so far has been used only for eye melanomas. In the future, modifications of the unit will allow therapy of intracranial targets when higher proton energies are available. In its final form, the proton therapy facility at TSL will harbour a second treatment unit. Here a rotating gantry for 200 MeV protons will provide a broad beam, which will enable treatment of tumours located anywhere in the body.

Key words: Proton therapy, eye melanomas, cyclotron.

The synchrocyclotron at the The Svedberg Laboratory (TSL, the former Gustaf Werner Institute) in Uppsala was previously used for biological experiments and radiotherapy with an external 185 MeV proton beam. During the period between 1957 and 1968, 69 patients were treated with a large-field, range-modulated proton beam (1–6). The proton beam was also used for narrow-beam irradiation of intracranial structures (7–9).

The cyclotron can presently (1990) operate with fixed frequency and proton energies up to 100 MeV. Later, the cyclotron is planned to operate with frequency modulation and yield proton energies up to 200 MeV (10). The proton energies of the fully reconstructed cyclotron are considered optimal for various clinical applications.

The proton therapy project at the The Svedberg Laboratory is a national project in which several Swedish university departments and hospital clinics are collaborating. The technical preparations were undertaken by staff from the Departments of Radiation Sciences, Oncology and Ophtalmology at Uppsala University, the Department of Hospital Physics at the Uppsala University Hospital, and the The Svedberg Laboratory.

The proton therapy facility at TSL will, in its final form, harbour two units: one narrow beam unit for treatment of eye tumours and small intracranial targets and one broad beam unit for more extended tumours elsewhere. The plan for the broad beam unit is to include a rotating gantry for 200 MeV protons (11). The narrow beam unit is at present used for eye melanoma treatments.

Malignant melanoma of the eye was chosen as the tumour type to be treated first since the energies presently available from the cyclotron are sufficient only for relatively superficial targets. Preparatory beam tests with 72 MeV protons started in Febuary, 1988. This energy was chosen for eye treatments as the range well covers all possible tumour localizations within the eye. The medical preparations for the eye melanoma treatments are carried out by the Departments of Ophtalmology and Oncology at the Uppsala University Hospital in collaboration with the Department of Ophtalmology at the Karolinska Hospital in Stockholm.

Later, in our treatment programme, the plans are to treat arterio-venous malformations in the brain and hormon active pituitary tumours. Higher proton energies are then necessary and will necessitate some modifications of the narrow beam unit.

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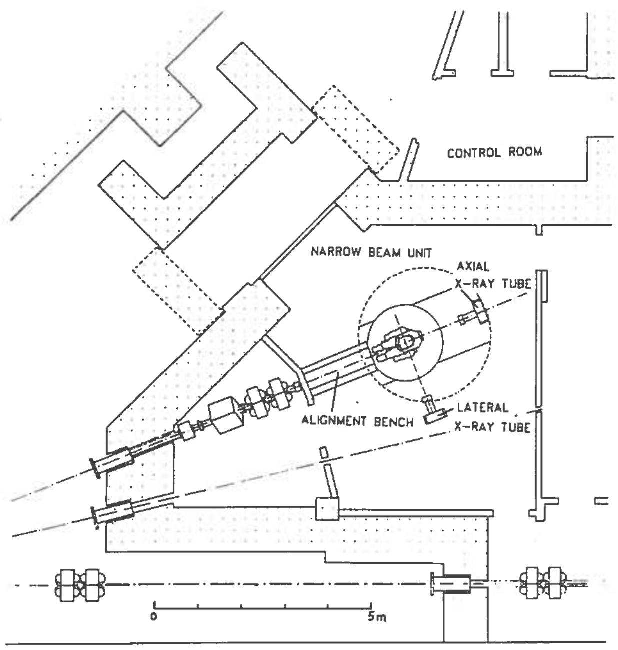

Fig. 1. Layout of the narrow proton beam treatment room.

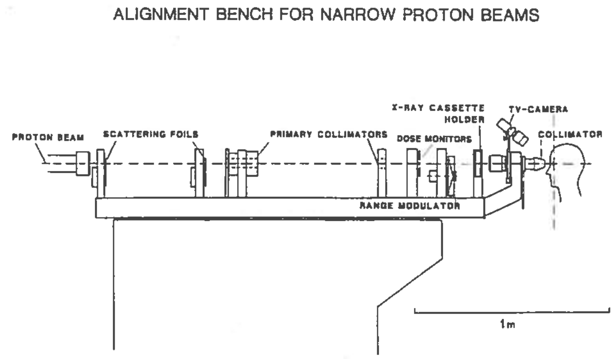

Fig. 2. The alignment bench for narrow proton beams.

We describe here the technical aspects of the narrow beam unit which is now in use for eye melanoma treatments. This type of treatment started at Harvard, Boston, USA (12). The basic ideas from Harvard have been taken up and further developed at Berkeley, CA, USA and at the Paul Scherrer Institute, Villigen, Switzerland. The design of this therapy unit has been influenced by the pioneering work at these centres.

The narrow proton beam treatment room

Fig. 1 shows the layout for the narrow beam unit. The unit provides a fixed horizontal beam at 1.5 m above the floor. The beam is transported from the cyclotron in a 65 m long vacuum line. An alignment bench is placed right at the end of the vacuum system. The bench contains elements for beam shaping and dose monitoring. The patient is seated in a movable chair for easy positioning, where the head of the patient is immobilized. The chair can be rotated and moved such that intracranial targets may be irradiated from any direction in the horizontal plane. Two x-ray tubes oriented perpendicularly to each other allow exact positioning of the patient. The parts of the narrow proton beam unit are described separately in the sections to follow.

Devices for positioning and beam shaping

Alignment bench

A scheme of the alignment bench is shown in Fig. 2. The total length of the bench from the vacuum window to the patient is 230 cm. After leaving the vacuum window (0.1 mm Mylar) the protons pass through two scattering foils which yield a good lateral dose homogeneity over a diameter of 5 cm. Two primary collimators absorb the protons scattered through large angles. Four silicon semiconductor detectors, placed symmetrically in a ring at the edge of the homogeneous part of the dose profile, are used as dose monitors. The Bragg-peak region of the depth dose distribution is extended in depth by allowing the protons to pass through a rotating absorber of varying thickness. An x-ray film cassette holder with a cross-wire centered at the beam axis is placed directly after the range modulator. The final beam aperture is defined by a brass collimator which is individually machined for each eye tumour. A device for eye gaze fixation during the irradiation is placed close to the final collimator. A small TV-camera attached to the final collimator holder is focused on the patient's eye to detect possible eye movement during the treatment.

Double foil scattering

There are basically two principles for lateral flattening of charged particle beams: by magnetic scanning of pencil beams or by scattering in metal foils. The advantage of magnetic scanning is that large homogeneous fields can be obtained without energy loss. The disadvantage is that scanning magnets with power supplies and control systems are expensive and complicated. When small fields are required, in our case up to 5 cm in diameter, the scattering technique is preferred because it is simple, reliable and inexpensive.

The large number of small angle deflections that occur when a charged particle beam passes through matter is referred to as multiple scattering. A theoretical description of this process has been given by Molière (13) and Nigam et al. (14). Elements of high atomic number are good scattering material as they have the largest scattering power and give the smallest energy loss. A single scattering foil produces a lateral fluence distribution of gaussian shape (Fig. 3). In order to obtain good homogeneity over a given diameter only the central part of the gaussian

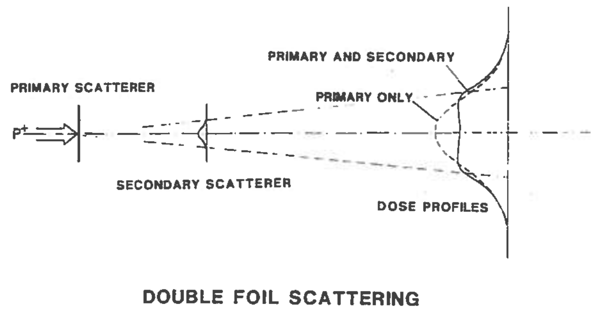

Fig. 3. The principle of double foil scattering. Indicated to the right are dose profiles obtained with the primary scatterer only and with the two foils together.

distribution can be used. The rest of the beam must be stopped by other means. Stopping of the peripheral beam particles induces unwanted nuclear activation and neutron production. The scattering technique can be improved by using double scattering foils.

The basic principle of double foil scattering is illustrated in Fig. 3. The idea is to spread the central part of the gaussian particle fluence distribution produced by the primary scatterer. This is done by placing a secondary scatterer of non-uniform thickness at some distance downstream from the primary scatterer. The profile of the secondary scatterer is almost gaussian. This technique has been used in clinical electron accelerators (15, 16). A theory for optimization of the thicknesses of the foils and the shape of the secondary foil was suggested by Brahme (16). We adapted the theory to handle proton beams and the mathematical details of this method will be described separately (Montelius et al. to be published).

The scatterers in our beam are made of brass, although lead or gold would be more efficient. The small field sizes required for our beam and the relatively low proton energy (72 MeV nominal energy from the accelerator) would require optimal lead or gold foils so thin, that they would be difficult to handle. The thicknesses of the two brass foils optimized for 72 MeV protons were 0.4 mm for the primary foil and 0.75 mm at the centre of the secondary foil. The lateral dose distribution measured with these two foils is shown in Fig. 4. A fairly rapid fall in dose outside the homogeneous region was obtained and the unwanted production of neutrons and radioactive isotopes in the stopping material on the alignment bench was minimized.

Range modulation

Fig. 5 shows a depth dose curve for a monoenergetic proton beam with a nominal accelerator energy of 72 MeV and a beam diameter of 5 cm. This curve was measured in a water phantom with a small semiconductor detector. The details of the dosimetry system are described below. As the lateral scattering of protons in tissue is small, the depth dose curve is practically independent of field size down to the smallest field diameters (7–8 mm) used for therapy. On their way to the phantom of the patient the protons lose energy in the vacuum window, the two scattering foils and the air. The total energy loss was calculated at 5 MeV and the depth dose curve thus corresponds to an energy of 67 MeV.

The narrow Bragg-peak must be spread over a larger depth in order to give a homogeneous dose to the tumour. Several techniques for modulating the range of proton beams have been reported in the literature (17–19). We have chosen to use a rotating, stepped absorber (18). The polystyrene absorber shifts the range in steps of 1.25 mm and an almost flat extended peak region was obtained

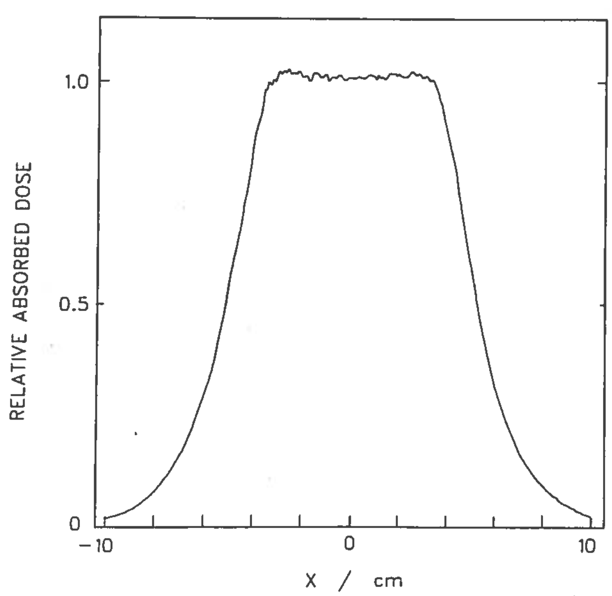

Fig. 4. Dose profile for a 67 MeV (72 MeV from accelerator) unmodulated, uncollimated proton beam. The dose is measured at 13 mm depth in water with a silicon diod.

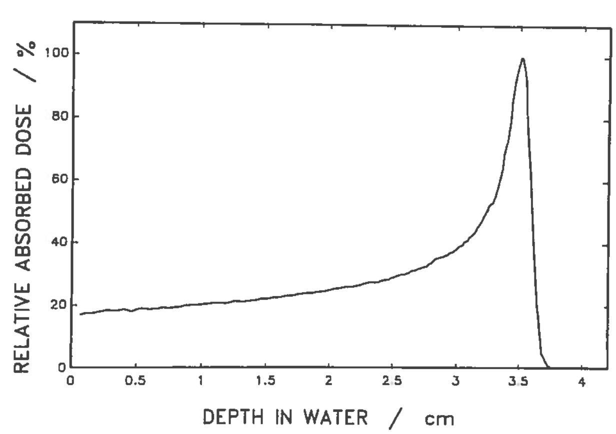

Fig. 5. Depth dose distribution of 67 MeV (72 MeV from accelerator) 5 cm diameter, unmodulated proton beam measured in water with a semiconductor detector.

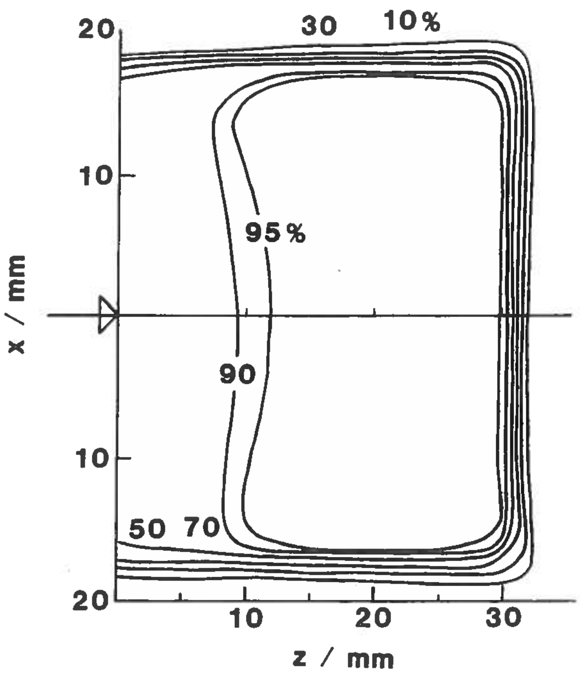

Fig. 6. Isodose distribution in water of a range modulated and collimated proton beam measured with an RFA-7 dose scanner. The extended peak region is 15 mm and the maximum range is in this case 32 mm.

(Fig. 6). Absorbers can be built for different ranges of the extended peak region. The method described by Koehler et al. (18) was used for calculating the shape of the stepped absorber.

The absorber rotates at a speed of 800 rpm which means that the Bragg-peak sweeps over the tumour a large number of times during one irradiation which normally lasts for 20–25 s.

Collimation system

The two primary collimators used to stop the protons scattered through large angles, are placed so that they protect the patient from unnecessary exposure. The final brass collimator is placed in a holder at the end of the alignment bench (see Fig. 2). The collimators for eye melanomas are machined individually for each patient. When larger fields are needed, the final collimator layout will be redesigned.

Coordinate system for eye fixation and tv monitoring of the eye

In order to deliver dose to an eye tumour with good geometrical precision, the patient's eye has to be in a fixed position during the treatment. To attain this position the patient is fixating a small light source positioned in a coordinate system. The coordinates of the light source are determined by the dose planning system. The coordinate system consists of an array of small lamps mounted on an arm which can be rotated around the beam axis. The polar lamp angle can be set at every second degree between 16 and 56 degrees and the lamp arm can be positioned at any azimuthal angle between 0 and 360 degrees. By using this lamp array, space is left free for a tv camera required for observing the patient during the treatment. The camera, a small CCD-type camera with high sensitivity and equipped with a tele-lens, is used so that small eye movements can be detected. While holding an oblique angle to the eye, the camera can be rotated around the beam axis so that the most favourable view of the patient's eye may be selected.

The treatment chair

The treatment chair (Fig. 7) was designed to be both flexible and very stable. It was designed not only for eye treatment but also so that any intracranial target can be irradiated from any direction in a horizontal plane. Three DC-motors move the chair in three orthogonal directions (x, y, z) so that any location within the head can be placed at the point where the beam axis and the rotation axis of the chair are crossing. The fixation device for the patient's head, designed primarily for eye tumour treatments, is also shown in Fig. 3. The fixation device is firmly attached to the iron bar located behind the patient. The immobilization of the head is based on a bite block which is a cast of the patient's teeth. Furthermore, the forehead support and neck supports attached to the fixation device can be adjusted to fit the individual patient. This fixation method is fairly simple and does not require fabrication of individual masks for each patient. The reproducibility from day to day of the position of the head is sufficient for eye treatments as the position is checked and adjusted before each treatment using radiographs.

For other intracranial targets a different fixation system might be preferred in which the diagnostic information from CT-scans is transformed to the coordinate system of the proton beam and treatment chair.

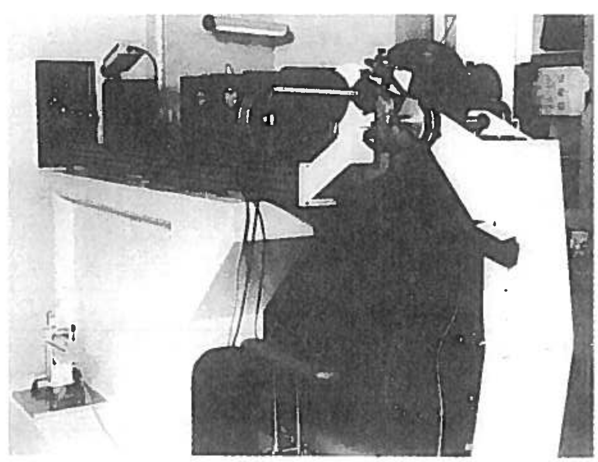

Fig. 7. Treatment unit with patient in treatment position.

X-ray image system for patient positioning

Two x-ray tubes are placed so that two perpendicularly oriented images of the patient may be obtained. The patient's eye is visualized on x-ray film by tantalum clips sutured on the surface of the sclera. The axial x-ray tube produces an image antiparallel to the beam axis (Fig. 1). The film is positioned so that the images of the collimator aperture and the eye clips can be seen. The lateral x-ray cassette holder is mounted from the ceiling and has a cross hair of thin tantalum wires indicating the position of the intersection of the rotation axis of the treatment chair and the axis of the proton beam. Polaroid film in modified cassettes is used in place of standard x-ray film. The image size is $9 \times 12\,cm^2$ which is large enough to cover the diameter of the eye. The advantage of this technique is the fact that the films can be developed quickly in the therapy room. This speeds up the patient positioning procedure in which a number of radiographs are taken until they are found to coincide with the images produced by the computer.

Dosimetry

Dose monitor system

The dosimetry and beam control unit, located in the control room, has two independent electrometers, both of which can switch off the proton beam when the accumulated charge in either of the electrometers reaches a preset value. The unit also contains an extensive safety and interlock system. The detectors used as dose monitors are silicon semiconductor detectors (Scanditronix P-Si field detector (19)). Four detectors are placed symmetrically in a ring outside the maximal collimator opening so that they do not disturb the useful beam. The signals from the vertical detectors are summed and read by one electrometer channel of the unit and the horizontal detectors are summed and read by the second channel.

In addition to the four dose monitor diodes, another array of four diodes is placed in the same ring in between the monitor diodes. They are connected to a four-channel electrometer which detects the differences in the signals between each of the four diodes. If the beam uniformity is so poor that the difference between any two of the diodes exceeds a preset value, the four-channel electrometer activates the interlock system and the beam is switched off.

Dose distribution measurements

Preclinical dosimetry tests. During preclinical tests, dose distributions were measured with a computer controlled dose scanner (RFA-7, Therados-Scanditronix). Small silicon semiconductor detectors (Scanditronix field-detector) were used to measure the proton dose distributions in a water phantom. In order to resolve the steep dose gradients, both in the depth dose distribution and in the sharp penumbra region, the size of the detector had to be small. Semiconductor detectors are ideal for these purposes as they can be made very small: the thickness of the sensitive layer was $50\,\mu m$ and the width of the detector 0.5 mm (20). Despite their small dimensions the semiconductor detectors gave adequate signals. During the tests the dose rate was kept rather low (< 10 Gy/min) in order to avoid saturation due to recombination. Results of the measurements are exemplified in Figs 4–6.

Clinical dosimetry tests. Before each treatment the dose distributions have to be verified. This precaution is considered necessary as the dose distribution in depth is formed individually for each tumour. By measuring the depth dose distribution, the correctness of the range modulation and maximum range of the beam can be confirmed. Likewise, the lateral homogeneity can be checked before each treatment. It is inconvenient to use a large water tank to measure the dose distributions before each treatment. Small scanners with solid absorbers, such as the depth dose scanner developed at the Paul Scherrer Institute (C. Perret, personal communication), are therefore used for quick dose measurements before each patient treatment. For the present application we have developed a small computer controlled scanner which can measure depth dose distributions down to 50 mm depth in tissue as well as lateral dose distributions for field sizes up to 50 mm (Fig. 7). This scanner uses a semiconductor detector similar to the one used with the RFA-7 scanner. The thickness of the absorbing material is varied by using a moving polystyrene wedge controlled by a stepper motor. Another stepper motor moves the detector sideways, at any desired depth, for measuring beam profiles. The movements are controlled by a computer which also registers the detector signal and displays the dose distributions. The dose curves measured by this scanner were calibrated against the curves measured by the RFA-7 scanner.

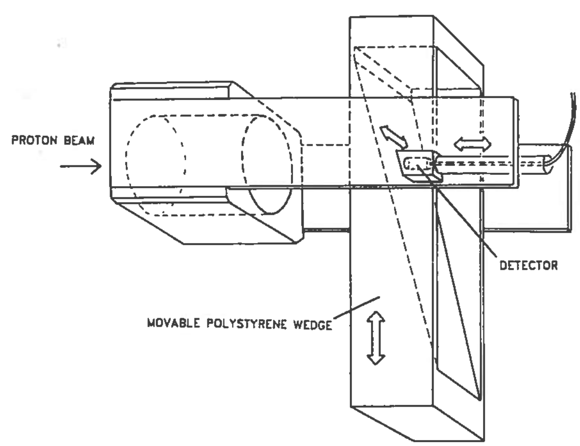

Fig. 8. Schematic view of dose scanner for quick tests of depth dose distributions and lateral dose profiles.

Determination of absorbed dose per monitor unit

The determination of absorbed dose to the tumour per monitor unit is based on dosimetry with an ionization chamber. This is one type of detector recommended by the AAPM protocol for heavy charged particle dosimetry (21). For this purpose a small, 0.1 cm^3, thimble type ionization chamber with tissue equivalent walls was used (Far West Technology, type IC-18). The chamber was used filled both with air and with methane-based tissue-equivalent gas. The chamber was calibrated in a ^{60}Co beam at the Swedish secondary standard laboratory in Stockholm. In the AAPM protocol it is recommended that the chamber be placed either at the plateau of the depth dose curve in a monoenergetic, unmodulated beam (Fig. 5) or in the extended peak region in a range modulated beam (Fig. 6). We have chosen to place the ionization chamber with its centre at 7 mm depth in the plateau of the unmodulated beam, where the absorbed dose per monitor unit is measure. In order to determine the absorbed dose in the tumour from the range modulated beam, we use a small semiconductor detector with a sensitive layer thickness less than 0.1 mm. It is placed both at tumour depth in the range-modulated beam and in the same position as the ionization chamber in the plateau of the unmodulated beam. This gives the relation between the tumour dose and the plateau dose earlier determined with the ionization chamber and thus is the tumour dose per monitor unit determined. The use of the small semiconductor detector instead of the ionization chamber in the modulated beam is motivated by the fact that, in certain cases, the extended peak region may be of the same size of smaller than the inner diameter of the ionization chamber, which is 8 mm. The perturbation of the proton beam caused by the ionization chamber would in such a case make the dose determination unreliable.

As this dose calibration is performed before each patient treatment, a polystyrene phantom has been built for rapid dose measurements. It can be attached to the holder for the bite block in the fixation device. This phantom can be used with both the ionization chamber and the semiconductor detector.

Dose planning system

In Boston, Goitein & Miller (22) developed a dose planning system specially designed for treatment of eye tumours. In this programme the eye can be projected from different directions. The programme uses the coordinates of the tantalum clips measured from two orthogonal x-ray images to create a computer image of the eye. With this programme the direction of the eye axis is determined in such a way that the proton beam will avoid sensitive parts of the eye, i.e. the lens and the optic nerve. Furthermore, the beam aperture, the maximum range and the range modulation of the proton beam is optimized. Dose distributions in different planes through the eye may be calculated and displayed. The programme also simulates x-ray images of the tantalum clips. These images are used to align the patient in the proton beam.

The group at the Paul Scherrer Institute (PSI) in Villigen has modified the program from Harvard. We have installed the PSI-version of the programme in a VAX-computer at the University Hospital in Uppsala. The programme has been adapted to local conditions. We have also installed a new routine which gives the percentage dose at any point within the eye.

The collimator is fabricated with a numerical milling machine which uses the coordinates for the beam aperture from the planning programme. Tests of the programme have been performed using an eye phantom. The programme can be run both from the hospital and from the The Svedberg Laboratory via a direct link. A dose planning station including a graphics terminal and a plotter is available at each of these locations.

Patient treatments

During 1989 and 1990, 19 patients with eye melanoma were treated. They were given 54.6 Gy in four fractions. Technically all treatments were successful. Our experiences will be accounted for separately.

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