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PROGRAM FOR A NOVEL ION GANTRY

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Abstract

In conjunction with the CERN-based development of a specially designed cancer therapy synchrotron (Proton-Ion Medical Machine Study – PIMMS), the need for the development of a facility equipped with an *ion gantry* emerged. Such a medical machine shall be capable of delivering a carbon ion beam virtually from every position and direction best suited for the treatment of the patient. Therefore the gantry allows to conform the delivered dose very closely to the tumour volume, minimise the dose deposited in healthy tissue and spare any critical organs. The technical challenges for the design are the high beam rigidity of the ion beam, requiring large and very heavy bending magnets, as well as the specified high precision of the whole system.

1 INTRODUCTION

New developments in radiotherapy are guided by the objective of conforming the delivered dose to the tumour volume as closely as possible. By the use of charged particles like protons and ions – compared to the classical approach of using photons – one can benefit from the so-called "Bragg-peak" effect: most of the energy of the particle is deposited at a certain depth inside the tumour, depending very accurately on the original energy of the particle.

An additional strategy to meet the above objective is to irradiate the patient from various angles and specified directions: the application of multiple fields reduces the dose delivered to the healthy tissue in the entrance channel and by using certain irradiation paths it is possible to spare any critical organs. Machines capable of delivering the required dose from every position and direction best suited for the treatment of the patient are called *medical gantries*.[1]

Such machines have been used for conventional and recently also for proton therapy.[2] Whereas for the classical radiotherapy the electron (linear) accelerators are light enough to be mounted directly on the gantry, this is not possible for protons (and of course not for the even heavier ions). The horizontally delivered proton beam has to be directed onto the patient by several bending magnets. Somehow this beamline has to be moved and rotated relative to the patient.

It seems desirable to provide the possibilities of a gantry also to the emerging and very promising application of ions in radiotherapy in order to show the full potential of this technology in direct comparison to proton therapy [1]. However, the rise in complexity of an ion gantry (compared to the already very complicated proton gantries) is considerable and the solution chosen will have a decisive impact on the financial efficiency of the whole facility. Simplistically, this increase of complexity is due to the higher rigidity of the proposed carbon ion beam compared to a proton beam (responsible for a dramatic increase of the corresponding magnet bending radii roughly by a factor of three.)

2 APPROACH

Theoretically, specific medical and beam-optical objectives and constraints guide the design of an ion gantry facility. However, as with every dynamic and evolving project, such specifications are very vague at the beginning. In fact, before making any decisions about the design of a facility, an (architectural) *program* is presented very briefly in this report focussing on the question: "*Which processes will happen in the future ion gantry facility?*" and "*What should be built in order to support best these intended processes?*"

As the aim of the ion gantry is to gather valuable – and so far non-existing – clinical experience, uncertainty about future operation scenarios is high and procedures might change

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[1] The original meaning of the word *gantry* refers to a moveable platform or crane.

[2] The first proton gantry was installed in the Loma Linda University Proton Treatment Center in the early 90'ies, two others followed. The Paul Scherrer Institut (PSI) started using their proton gantry in 1996. The Northeast Proton Therapy Center (NPTC), Boston, and the National Cancer Center, Kashiwa, are going to commission two proton gantries each during 1999.

substantially with growing number of treated patients due to promising results. This perspective leads to the question *“What could be needed in the future ?”*

When answering the above question in a structured and documented decision making process, principal systems of the ion gantry are elaborated and possible technical solutions generated.

3 USER, INVESTOR, OBJECTIVES

The first task of the programming-stage is to identify investors and future users of the facility and their individual expectations and objectives, which might very well be contradictory sometimes.

The novelty of the ion therapy in general and specifically the ion gantry will call for a public funding. Therefore the objective of the investor can be defined as

- Investment into a novel health-care facility for ion-therapy using a gantry, which needs as little financial support as possible
- Investment into a world leading research facility (“centre of excellence”).

The latter implies that the community of users does not only consist of patients and staff, but also scientists. Additionally, the number of staff members will be higher compared to a conventional radiotherapy unit. The designated facility must provide the necessary resources (laboratories, conference rooms, “think cells”, etc.) but apart from that the built surrounding should enhance formal and informal communication and therefore stimulate information exchange, the joint development of new knowledge and creativity.

Patients are (weak) customers judging the quality of the treatment by the building they see and the people they meet. A holistic view requires the facility not only to fulfil technical specifications but also to mitigate the patients fears, encourage his self-confidence, dignity and will to overcome the illness and give him guidance, in order to positively influence the patients physical health. To support these objectives the whole facility shall provide:

- Simple, understandable and logically arranged procedures (short distances, compact design, transparency, communicative surrounding, information systems etc.)
- Comfortable and reassuring impression (no bunker-like building, daylight, generous room geometry, colours, natural ventilation where possible – no “hospital smell”, discreet access controls and safety measurements, respect of patients privacy etc.)
- Layout favouring personal contact between personnel and patients

Ion-therapy is a major research topic and rapid scientific progress can be expected in the future. The new ion gantry facility has to assimilate and even support these developments making internal (variability) and external (expansion) flexibility mandatory. Additionally, (medical) specifications are more than vague and time from first planning steps until operation is very long.

Therefore, flexibility for changing demands and new techniques has to be provided concerning

- patient throughput,
- equipment,

- research activities.

4 FUNCTIONAL SPACE–PROGRAM

4.1 Patient Flow and Activities

An ambulant patient coming in for having a regular treatment session, i.e. a fracture, will have himself announced at the **reception** and wait in the **waiting and information area** until he is called for treatment. The patient proceeds to one of the three **dressing rooms** (Fig. 1). A **sub-waiting area** also functions as the waiting area for accompanying persons. **Toilet facilities** shall be adjacent.

The lightly dressed patient will then see the physician in an **examination room** or directly proceed to the **gantry room** (patient cabin). The set-up procedure requires the patient to lay down on his specifically prepared mould (temporarily mounted to the patient positioning system - PPS) in order to get immobilised.[3]

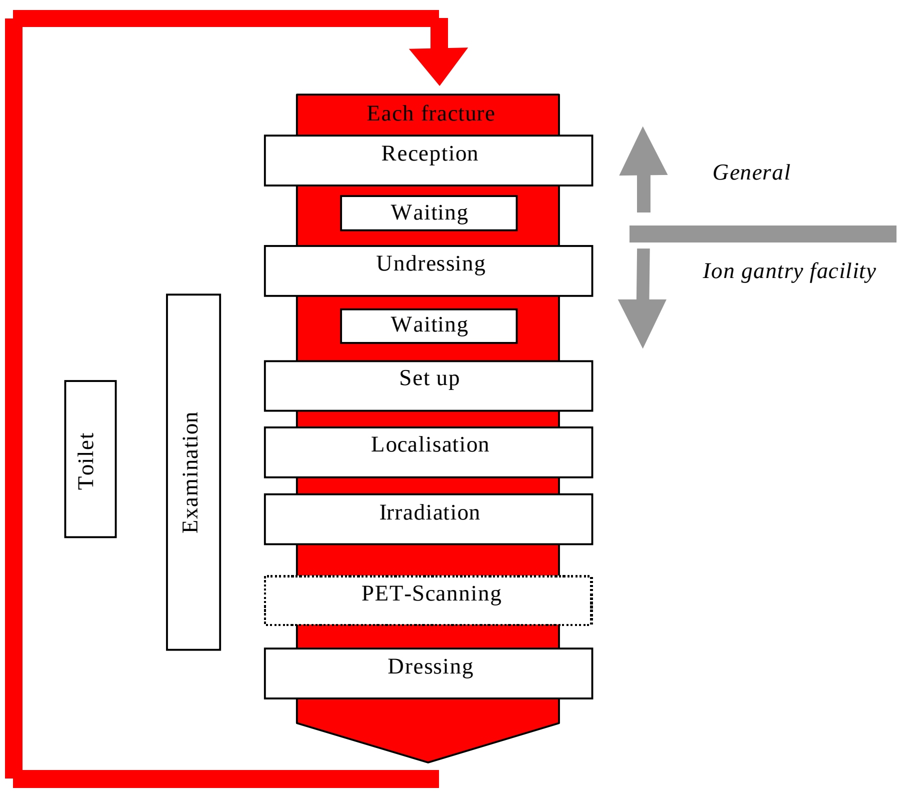

Figure 1: The process “fracture” from the patient point of view.

X-ray images in horizontal and vertical planes, and / or in the direction of the beam will be taken for exact localisation of the target towards the actual isocentre of the ion beam. The digitised images will be displayed on a screen and manually compared to the DRR (digitally reconstructed radiograph, i.e. basically a 3D patient model reconstructed from the CT’s on

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[3] When irradiating head and neck tumour, individual masks will be used instead of the mould.

which the treatment planning is based). Necessary adjustments to the patient's position are applied and a physician performs a final check.

The whole localisation procedure lasts for about 10 to 20 minutes, the actual irradiation for about 2 min. Only the patient is allowed to be in the gantry room during irradiation, which is supervised by the therapists from the adjacent **gantry control room**.

Irradiation with carbon ions offers the possibility of an instant in vivo dose localisation during or immediately after irradiation with a PET-camera (Positron Emission Tomography). However, the merits and the actual procedures required by this technology are still a matter of discussions among experts.[4]

Principally, such a PET camera could be mounted directly on the gantry or installed in a designated PET-room. If the camera is fixed to the gantry and taking the image lasts longer than the irradiation, then the gantry is blocked while the PET image is taken, which results in a reduced patient throughput of the facility. Additionally, a hadron facility would certainly have another ion treatment room (probably with a fixed beam) for which the use of the PET would be desirable too. On the other hand it has not been studied yet, in which way a transfer of the patient from the gantry room to the PET room immediately after the irradiation would affect the image-quality adversely.

INPATIENTS

The design of the facility has to support treatment of inpatients. Space has to be reserved for a designated **inpatient waiting area**, which should provide some basic medical utilities and direct visual contact to a staff member. Relevant rooms and circulation areas have to accommodate rolling stretchers.

4.2 Material Flow

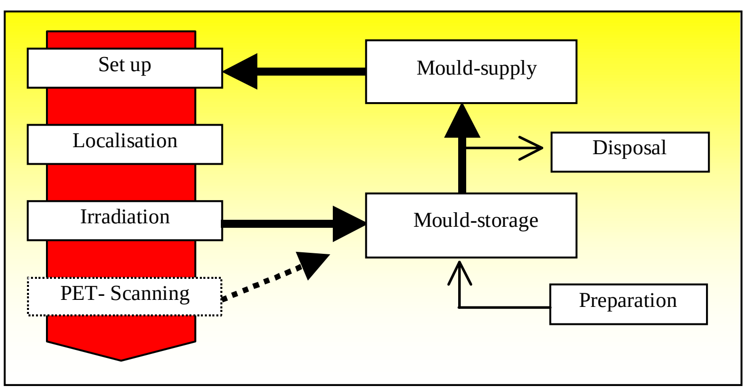

Figure 2: The major material flow in the ion gantry facility: supply and storage of the moulds (pods).

The only material-flow relevant from a logistical point of view concerns the supply and storage of the moulds (pods). Initially, the intended patient throughput of 2 patients per hour[5] requires storage capacity for at least 35 of these fairly large items per gantry. For every treatment the right mould has to be supplied for the immobilisation of the patient and removed

[4] So far, the successful simultaneous control of irradiation with a PET-camera was confined to head and neck treatments [2].

[5] First approximation: operation 15 hours per day, 300 days per year; 22 fractions per patient, 2 fractions per hour, giving 9000 fractions equalling 410 patients per year.

to the storage area, after the treatment. Therefore, the **mould-storage room** must be close to the gantry room.

For the purpose of quality checks and quality assurance there will be a dedicated room, where certain devices are stored and a small workplace for a physicist is provided (**quality assurance room**).

4.3 Outlook on Possible Future Operation Changes

AUTOMATION OF THE LOCALISATION PROCEDURE

Of the total time the patient occupies the gantry room, more than 50 % is used for the alignment check and correction, i.e. the "localisation". Automatic comparison between the digital images and the DRR would significantly speed up the treatment and increase the utilisation of the gantry. Depending on the degree of automation, the scheduled 25 minutes per patient could be reduced to approximately 15 minutes (resulting in an increased demand for mould-storage and dressing facilities).[6]

AUTOMATION OF ALIGNMENT

If somehow the localisation process is shifted to a separate "**preparation room"**, performed less often, or even becomes unnecessary, the patient occupancy of the gantry room will be reduced even further.[7] In principle, the location of the tumour has to be defined relative to an accurate local reference system (e.g. an immobilisation device, on which the patient can be fixed in a reproducible way). This system and the actual isocentre of the beam (gantry) must then be related in a co-ordinate way (mechanical connection, optical recognition, etc). However, limited beam availability due to accelerator occupation will probably prevent a further increase of the patient throughput.[8] Especially in case of multiple field irradiation with ions or respiration triggering becoming common practice it is very probably that the beam availability (and not the duration of the actual activities taking place in the gantry room) *will* limit the throughput in the ion gantry.

POSITRON EMISSION TOMOGRAPHY (PET)

Research concerning the technical features and the potential fields of PET-utilisation progresses very rapidly. In-vivo dose localisation and quantification seem promising fields of application. Therefore the facility and the gantry should be capable of supporting intensive use of PET imaging.

4.4 Functional Relationship and Space Requirements

Figure 3 shows the functional incorporation of the ion gantry unit into the hadrontherapy facility and represents a sound basis for preliminary architectural planning.

Quantity (area) and quality requirements (standards, adjacencies) for every room will be assigned and specified during the design phase. Special attention has to be paid to the

[6] IBA for example calculate the gantry room occupation time for the scenario outlined above to be 12,5 min [3].

[7] Compare for example the proposals presented in reference [4] stating that for automated patient alignment two preparation rooms should be provided per treatment room.

[8] In a hadrontherapy centre with 5 treatment rooms and a throughput of 4 patients per hour and room, the "window of beam availability" per patient would already be limited to 3 min.

possibility of installing, removing and exchanging the heavy and bulky equipment inside the gantry room. Considerable space for the power supplies of the gantry must be provided in the switchyard or in a separate room.

A **space reserve** shall be foreseen adjacent to the gantry room in case a PET room, preparation room or re-mobilisation zone is needed in the future.

The number of dressing rooms, toilet facilities and the capacity of the mould storage must be easily adaptable to an increased patient throughput.

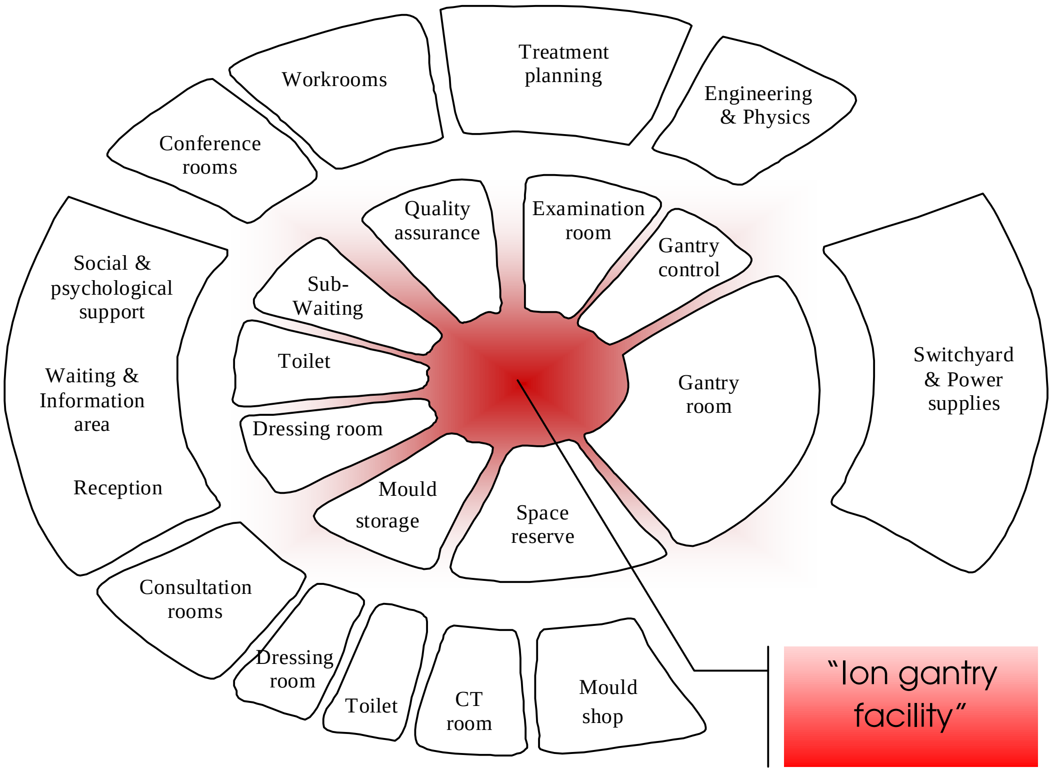

Figure 3: The ion gantry facility unit embedded in a hadrontherapy facility. Operation scenarios and various flows (patient, staff, information and material) can be superposed and checked.

5 FUNCTION OF THE ION GANTRY

5.1 Strategic Objectives

To focus the efforts, the development of an ion gantry is guided by four strategic objectives:

SAFETY- SIMPLICITY OF DESIGN AND OPERATION

The ion gantry will operate and prove its clinical advantages in a hospital environment; consequently, a cautious and conservative approach to new technical developments is mandatory. Complexity of design and operation procedures must be kept to a minimum to avoid operational errors. A routine day-to-day clinical operation requires reliable and not over-engineered solutions, which are understandable and easy to handle for the staff.

FLEXIBILITY

Continuously, clinical experience and research results as well as new technical developments will have to be incorporated into the design and the operation of the ion gantry. This requires a design to consist of open and flexible systems.

QUALITY

Two crucial issues drive the quality of the treatment:

- The technical performance, i.e. the precision of the treatment, the available beam directions towards the patient, etc.
- Human factors and ergonomics concerning patient and staff e.g. cabin layout, atmosphere, lights, comfort, etc.

EFFICIENCY

Treatment with the ion gantry based in a clinical facility must be competitive with other therapies (relative to its merits). Therefore cost efficiency with respect to facility life cycle costs and daily operation (cost of necessary staff) is a strategic objective.

5.2 Specifications

The technical specifications presented in table 1 are the result of the previous chapters and the strategic objectives. They represent the basis for the development of the ion gantry.

SAFETY FLEXIBILITY QUALITY EFFICIENCY

Specifications for the ion gantry		
Patient position	Supine to restrict organ movements to a minimum.	Supine
Irradiation sites	It has to be assumed that treated indications will comprise all regions of the human body.	All regions
Treatment angles	The author found no statistics about treatment angles actually used in routine clinical operation. Full 4π-irradiation to a patient in a supine position is the optimum to achieve (gaining a maximum of flexibility). Limiting or even blocking the rotation of the patient positioning system (PPS) around the vertical axis (corresponding to a transformation from 4π to 2π-irradiation) would result in a slight reduction of the overall gantry-dimensions and the complexity of the PPS only.	4π-irradiation
Field size	No medical studies about the statistical distribution of field sizes in radiotherapy are available. Symmetric field sizes are preferred for flexibility reasons. The maximum extension of a tumour can be as large as 40 cm. From the medical point of view minimal field seizes of 25 x 25 cm^2 are desirable [5, 6]. However, it is expected that this parameter will have a decisive impact on the design, the overall dimensions and the cost of an ion gantry. Therefore, economically reasonable values will be in the region of 15 x 15 cm^2 to 20 x 20 cm^2.	> 15 x 15 cm^2
SAD[9] / SSD	Considering a point source of the beam, a decrease in the effective SAD (or equivalent SSD) leads to an over-proportional increase of the relative dose on the patient surface (skin); the skin dose to target dose ratio decreases with an SAD-increase, making a large SAD desirable. Usually, an effective SAD larger than 2 m is considered as acceptable, but 3 m are highly preferred. [10] [5, 6]	Effective SAD $\geq$ 2,5 m
Spreading / Scanning system	Precise conformal tumour treatment offering maximum flexibility in treatment planning requires the ion beam to be actively directed ("scanned") over the arbitrarily shaped target area. No patient-specific hardware is necessary (except from the mould) allowing multiple field treatment in one single session without intervention of the personnel.	(Active) Scanning
Beam rigidity	Reaching deep-seated tumours requires the carbon beam to have a beam rigidity of at least 6,3 Tm (corresponding to a penetration depth of 27 cm in water). A normal conducting bending magnet having a magnetic field of 1,8 T would need to have a radius of 3,5 m.	~6,3 Tm
Positioning accuracy	It is assumed that the ion gantry is (also) used for high precision irradiation requiring sub-millimetre accuracy (e.g. brain tumours). If reasonably feasible, the tolerance for the technical system ion gantry (alignment and irradiation) shall be ± 0,5 mm.	Tolerance: ± 0,5 mm

Table 1: Medical and beam optical specifications for the ion gantry.

[9] The *source to axis distance* (SAD) is qualitatively similar to the *source to surface distance* (SSD). The terminus SAD refers originally to isocentric proton gantries and specifies the distance between the first beam scatterer ("source") and the isocentre. In the context of other gantry principles one should use the expression "effective SAD", which can be defined more generally as the distance between the (possibly virtual) point source of the beam and the "local" isocentre where the Bragg-peak occurs.

[10] In the extreme case of having 30 cm between the irradiated slice of the tumour and the skin, a 2 m and 3 m effective SAD is responsible for a relative (!) dose increase on the skin of 39 % and 23 % respectively.

6 SYSTEMS

6.1 Emergency Access

It has to be assumed that the patient is unable to get himself out of the mould without any help from another person.

It is very probable that any emergency situation will first be realised by a staff member present in the control room. As in conventional radiotherapy, this person has to leave the control room, pass through the maze (labyrinth) and enter the gantry to save the patient. This procedure lasts for about 10 to 15 seconds.

In a gantry where the patient is virtually kept at the same position all time during treatment (isocentric gantry), direct access to the patient is automatically possible at *any* time.

In a non-isocentric gantry, where the patient is moved considerably up and down during the session, these 15 seconds can be used to reliably move the patient to a position where direct access is possible ("access-position"). Several emergency scenarios for non-isocentric ion gantries and their resulting technical requirements are highlighted in table 2.

Emergency situation	Hazard prevention	Action to be taken
General alarm		Gantry is moved to access-position* and the patient is rescued
Patient hazard	❑ Presence of personnel until right before irradiation starts ❑ Comfortable mould ❑ Short duration of the session ❑ Smooth and quiet mechanical systems	Gantry is moved to access-position to take necessary measures
Gantry breakdown	❑ Reliability and simplicity ❑ Preventive maintenance	Access to and rescue of the patient via an independent second access system
Fire alarm in the gantry room – rapid evacuation	❑ Minimise usage of combustible materials ❑ Fire and gas detectors, smoke ventilation	Redundant, fire safe emergency drive system to move the gantry to access-position and rescue the patient

Table 2: Possible emergency situations in a non-isocentric gantry and their implications for the design. (*) "Access-position": the reference position where free (horizontal) access and exit to the actual treatment cabin and the PPS is possible.

In order to guarantee quick access to the patient wherever the cabin of a non-isocentric gantry is, *two* independent, non-interfering access systems have to be provided; one of which is the gantry itself (redundancy). This second system can be some kind of robotic arm, elevator, turning wheel, etc. Preferably it should be an already widely used standard system with only minor modifications.

For the unlikely case of hazard combination a third, "conventional" rescue system shall be foreseen (ladders, crane, inflating slide, etc.).

6.2 Beam-optics

GENERAL

To deliver an actively scanned, vertical ion beam to a supine patient from any arbitrarily chosen direction, the minimum requirement is to have one 90° bending magnet (dipole) in conjunction with a PPS, which rotates around a vertical axis. Several quadrupoles will be used for beam focussing.

Usually, alignment for the magnets is done with a precision of ± 0,1 mm [7].

The influence of a gantry structure made of (conventional) steel on the magnetic field has to be checked. Possibly, parts of the gantry structure close to the magnets have to be made of (expensive) austenite stainless steel, which has non-magnetic characteristics.

For accelerating the carbon ions it is assumed that a *synchrotron* will be available as presented in the proton-ion medical machine study (PIMMS) using *slow resonant extraction* [8]. The design value for the extraction energy is 120 – 400 MeV per nucleon, the number of carbon ions per spill is 4 x 10^8. Time for a nominal treatment is calculated to be 2 min (60 spills of 1 s plus 1 s to ramp up and down), sufficient to deliver a nominal dose of 2 Gray to a two litre volume.

SCANNING

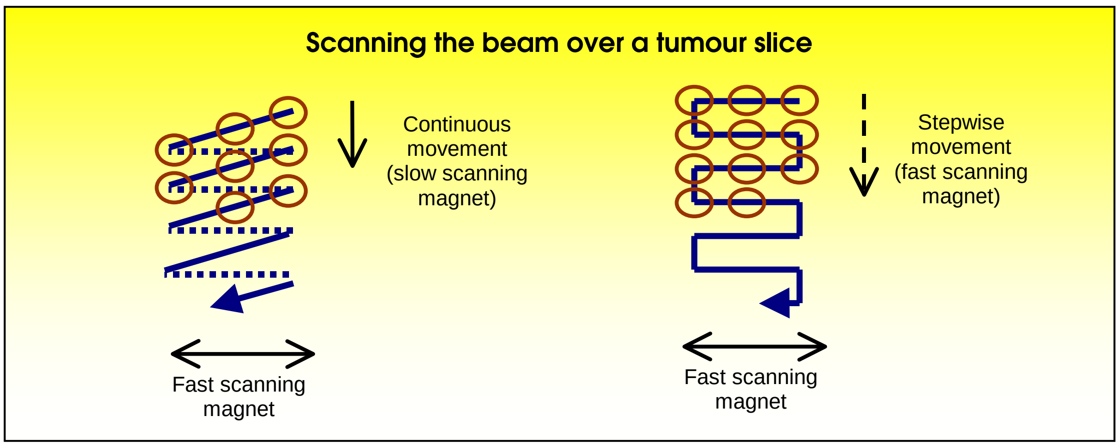

Figure 4: Impact of the scanning method on the necessary performance of the scanning magnets.

The principle of scanning is to divide the arbitrarily shaped tumour into small volume elements (mm-region) and irradiate each of those elements separately using a pencil shaped beam. This ion beam has to be directed very rapidly over the tumour volume, requiring *three degrees of freedom:*

- The depth is varied by *energy variation done by the synchrotron* every cycle (~2 sec.). This represents the slowest dimension of motion. The tumour will be irradiated slice by slice, the slices being perpendicular to the direction of the beam.
- A *scanning* magnet performs the fastest movement.
- The remaining third dimension should direct the beam over the whole diameter of a slice within a cycle of the synchrotron. Performing this movement continuously would require a minimum speed of ~20 cm/s and the beam would have to “jump” back very rapidly after a line of spots is finished (Figure 4, left). However, with the very inhomogeneous irradiation patterns expected for a tumour slice this procedure would suffer from considerable dead times. Therefore, stepwise movement for the third dimension is preferred. In an extreme case where the beam is not switched off during movement the required velocity for both scanning directions would theoretically be the same (Figure 4, right).

The considerations above suggest that – in case the energy variation is done by the synchrotron – scanning in the two other dimensions should to be performed by scanning magnets. Flexibility in treatment planning and shorter irradiation times favour similar scanning velocities for both dimensions. Consequently, solutions where the cabin or the last bending magnet itself [9] actively provide a scanning dimension seem unfavourable. A movement of the PPS cannot accomplish even the minimum speed for the second dimension.[11]

Figure 5 indicates that basically the scanning magnets can be placed before (upstream) or after the last bending dipole (downstream), having considerable effects on the weight, seize and cost of that magnet and the overall dimensions of the gantry and the building.

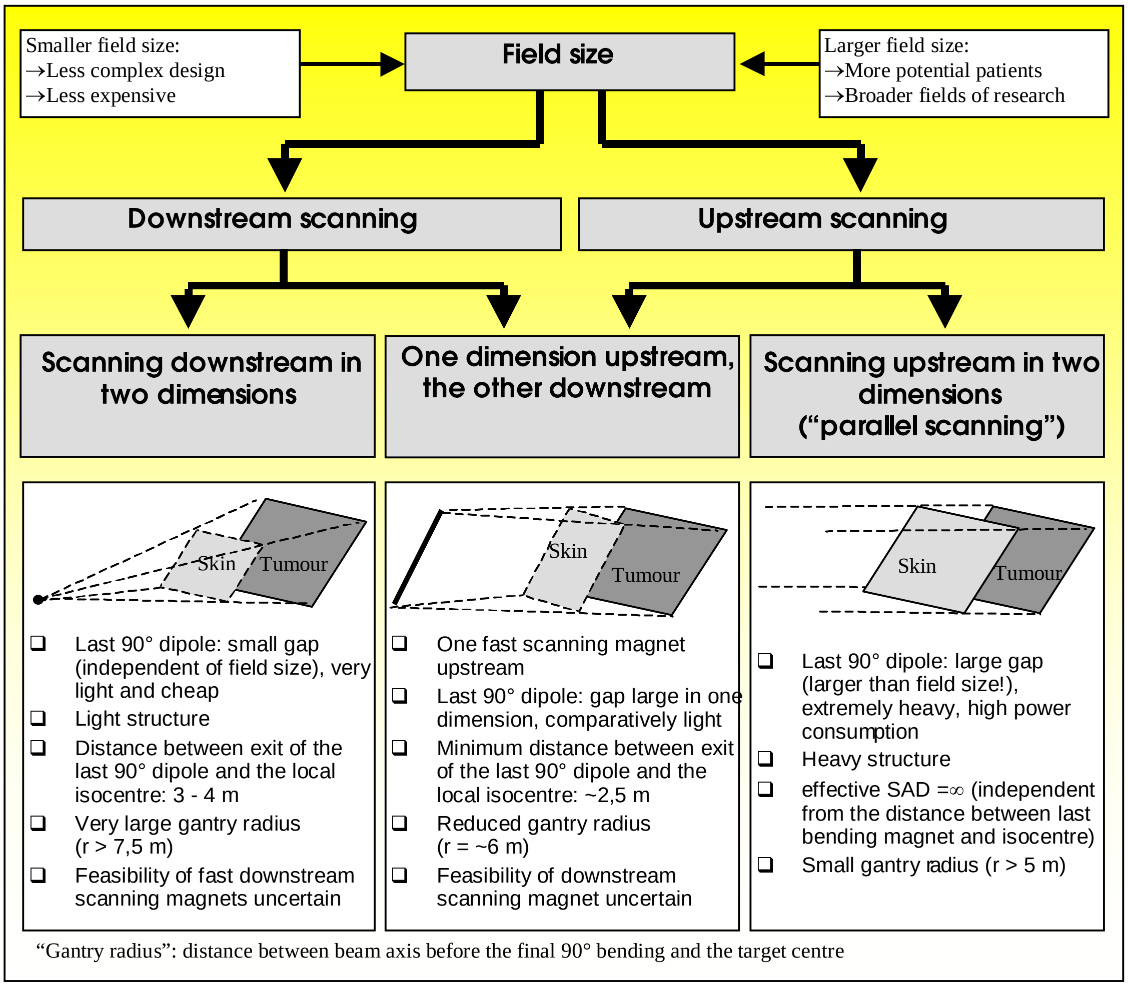

Figure 5: Principal possibilities of arranging the scanning magnets to irradiate a certain field size.

[11] The proton gantry at the Paul Scherrer Institut (PSI) uses a fast range shifter for energy variation, which is mounted in the nozzle (where also the beam monitoring system is situated). Velocity is high enough to cater for the second scanning dimension, consequently the slowest motion can be performed by the PPS.

BENDING MAGNETS

Regarding a conventional, i.e. non-superconducting magnet, its developed magnetic flux density (magnetic induction) for an applied magnetic intensity can be seen as being composed of two components:

- The original part due to the powered coils
- An additional part due to the facilitating ("field supporting") effect of a ferrous core.

The power consumption of the dipoles is proportional to their magnetic reluctance. This reluctance can be reduced by decreasing the gap of the magnet or by improving the magnetic permeability of the iron core. The latter is done on one hand by keeping the required magnetic flux well below saturation of the iron core[12] – 1,8 T seems to be a reasonable value – and on the other hand by providing a large cross-sectional area of the iron core. Certainly, both these measures have adverse effects on the gantry structure, as the radius and the weight of the magnets are increased respectively.

It is obvious that to a certain degree the power to drive the magnet can be reduced by placing more iron on the magnet (Figure 6).

Very generally, the cost of a magnet is proportional to its weight, the cost of the power supplies rises proportionally with the installed power.

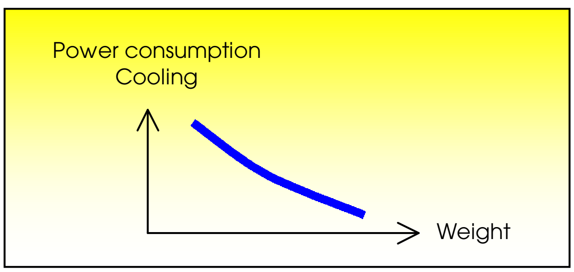

Figure 6: A ferrous core in the bending magnet "supports" the magnetic flux, the necessary power to drive the magnet can be reduced. Low power solutions give less trouble with the cooling and ventilation than high power magnets.

Scanning in front of the last 90° bending magnet ("upstream scanning") reduces the gantry radius but goes along with the need to enlarge the gap of this magnet – at least to the minimum field size (> 15 x 15 cm^2). A conventional iron-based magnet of this type is estimated to weigh between 60 t and 80 t and to have an overall maximum power consumption of a few hundred kW.

The alternative is to reduce the amount of iron in the magnet as suggested in reference [10]. However, the price one has to pay for this 40 t low-weight, wide aperture magnet is the enormous power consumption of 720 kW for 90° of bending and the related cooling difficulties (see below).

In comparison to these wide aperture magnets one has to regard a conventional magnet used in case of "downstream scanning" with a gap of a few centimetres only (in both directions). Weight and power consumption would be low, but unfortunately a much larger gantry radius would be needed to respond to the necessary effective SAD as well as to insert the downstream scanning magnets.

A solution to combine the advantages of the two principles – upstream *and* downstream scanning – is to place only one scanning dimension downstream: weight and power consumption of the bending dipole can be kept comparatively low, which facilitates gantry

[12] As long as the iron core is unsaturated, its permeability is manifold the one of free space, when saturation commences the permeability decreases continuously to meet the value of free space asymptotically.

and ventilation design[13]. The gantry radius is moderately enlarged to accommodate the downstream scanning magnet and to provide the necessary “SAD” (less than 2 m [14]). However, necessary velocity and high divergence angle of the downstream scanning magnet pose severe challenges to the magnet design.

COOLING AND VENTILATION

In order to get a stable magnetic field, uniform cooling of the magnet is essential. Additionally, thermal deformation of the system would jeopardise its high accuracy and must be avoided.

Basically, cooling of magnets is achieved by water running through them (with a delta T = ~20°). However, about 10 % of the total power are directly emitted to the air.

Cooling of high power magnets would therefore require the installation of heavy (and expensive) cooling equipment. Furthermore, the avoidance of temperature gradients during the short irradiation period demands for extremely powerful ventilation in the gantry room causing problems with dust, noise, etc., but still it has to be doubted that the accuracy of the system during irradiation can be guaranteed. To alleviate this problem one could increase the thermal mass of the steel structure by filling part of it with concrete.

SUPERCONDUCTING (SC) MAGNETS

Superconductivity offers the possibility of reducing the radius and in particular the weight of the bending magnets; a more compact and light gantry design would be feasible.

Cooling devices could be connected to the gantry by flexible pipes.

However, a lot of development for the intended use of SC-magnets would be necessary, addressing especially the following crucial questions:

- Which amount of iron is needed to avoid harmful magnetic stray fields acting on the beam monitoring equipment, the scanning magnets and the patient? Does this jeopardise the original saving in weight?
- Assuming the same effort, which field-quality can be achieved compared to a conventional magnet?
- Quenching of a SC magnet, i.e. a kind of mini-explosion, may happen. What effects does such an event have on the close patient and the accuracy of the system?

An advantageous compromise could be the use of magnet with a “warm” iron core plus superconducting coils.

Regarding the strategic objective of safety and reliability, it has to be said that superconductivity and its complicated cooling systems would add a critical technology to the (already) complex system of an ion gantry.

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[13] To get an idea of the feasible dimensions: a design presented in reference [11] suggests a weight of 22 t and a power consumption of 13 kW for a 90° dipole with a gap of 26 x 200 mm^2.

[14] Because the fast scanning is already done upstream, the downstream “source” of the beam can be regarded as a line and not as a point source (Figure 5). In the extreme case of having 30 cm between the irradiated slice of the tumour and the skin, a 2 m distance is responsible for a relative dose increase on the skin of 15 % corresponding to an effective SAD of 3,84 m.

6.3 Treatment Precision

The precision of the tumour irradiation is affected by various inherent, systematic and random errors due to

- the imaging systems, diagnostics and the ongoing treatment planning [12]
- the patient himself (organ movements)
- the beam generation and the beam optics as well as the lateral scattering and fragmentation of the ion beam
- the PPS
- the mechanical gantry structure.

This study deals mainly with the last items, however, it is essential that one always keeps in mind the whole system and the relative contributions of each part in order to identify the most suited areas of improvement.

The mechanical gantry structure, the PPS and the beam optics are highly coupled subsystems. Jointly they are responsible for delivering the beam to the patient with the specified absolute accuracy of ± 0,5 mm.

A tolerance of ± 0,5 mm (absolute accuracy) for the system "ion gantry" corresponds to a relative accuracy (relating to the overall dimensions of the gantry) of less than $1*10^{-4}$. This value calls for the introduction of active correction mechanisms.

There is the principal possibility that one of the three subsystems involved corrects errors made by the others, for example the PPS or additional correction magnets can be used to compensate deflections of the structure. However, this opportunity is restricted to inherent errors (e.g. elastic deflection) and systematic errors (e.g. uneven roller surface) only[15]. Random errors, i.e. statistically distributed errors (e.g. backlash), will have to be added geometrically, therefore decreasing the remaining "error budget" with each subsystem added.[16]

Excessive use of correction algorithms and devices will considerably increase the complexity of the system as well as the installation and testing procedures. Having the objective of reliability in mind, a high *initial* accuracy and rigidity of the mechanical gantry structure shall be guaranteed.

6.4 The Patient Cabin

The patient cabin is the apparent "room" where irradiation is happening. Its atmosphere and enclosure should make a smart impression to the patient. The PPS operating envelope seizes the minimum inner dimensions of the cabin. Space to manoeuvre with rolling stretchers and to load and unload the patient has to be guaranteed, as well as the possibility of walking around and shortly examining the patient being already in his treatment position.

Sufficient load capacity must be provided for the PPS[17], the x-ray equipment, several staff members and potential future equipment (e.g. PET-camera). The Cabin entrance must be wide enough to be entered by stretchers or similar moveable positioning devices.

No storage capacity for the stretchers is foreseen, as they will be stored in the maze.

[15] Inherent and systematic errors are also called repeatable errors and can be avoided; they are added arithmetically.

[16] Compare the error budget calculation in reference [13].

[17] The weight of the PPS installed in the NPTC is around 2,5 t.

If the cabin also provides some x-ray shielding and a terminal for the localisation procedure, then no additional work zone area will be necessary in front of the gantry, the entry will be directly from the maze.

TWO CABINS

To speed up treatment (without increasing personnel proportionally) it is principally possible to set up and prepare one patient while another one is irradiated. This preparation of a second patient would have to happen outside the gantry room to guarantee radiation protection for staff members. Ways of achieving this is to provide two cabins which are alternatively filled with a patient already prepared on its (movable) PPS, or to move in and out two separate cabins.

These long-term perspectives should be kept in mind when designing the – possibly flexible – front wall and the maze.

6.5 Gantry Structure

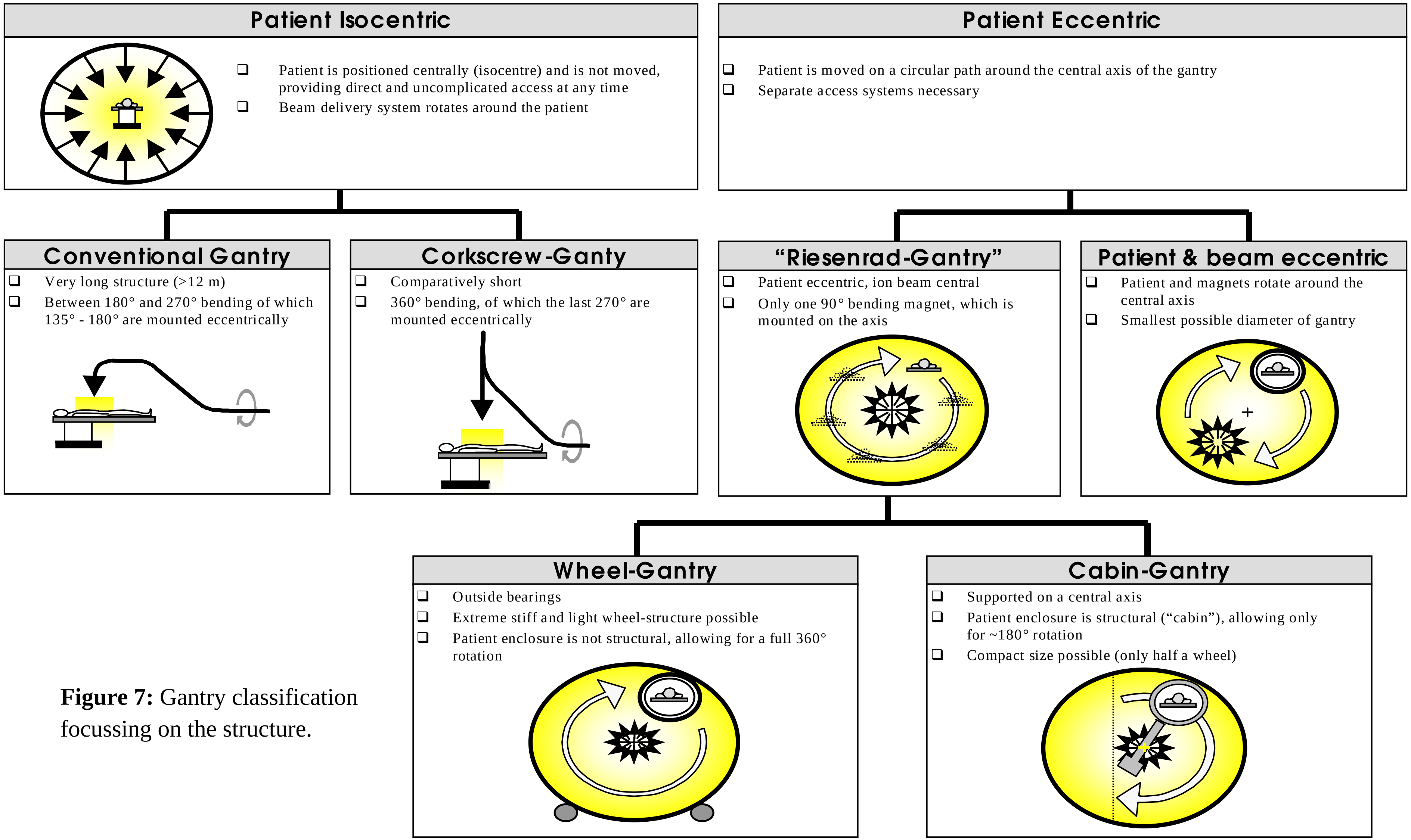

Figure 7: Gantry classification focussing on the structure.

Figure 7 shows several different structural approaches to an ion gantry. In the following paragraphs the different systems and their advantages will be presented.

ISOCENTRIC GANTRIES

Systems where the patient is placed in the centre of the machine are called isocentric gantries. All the proton gantries except from the one at the Paul Scherrer Institut (PSI) have so far been based on this approach.[18] The PPS is mounted for instance on the floor of the building and cantilevers into the gantry.

The incoming beam is guided away from the axis and eventually bent down towards the patient. This whole delivery system rotates around the centre **(conventional gantry)**. A special version of this is the **corkscrew-gantry**, where the beam is bent not only in one, but in two planes, giving a shorter construction length. However, about 315° of the total necessary 360° of the heavy bending magnets have to be mounted eccentrically, therefore not favouring an efficient structural solution in the case of carbon ions.

The construction of a conventional gantry for ions would imply to have a 12 m to 15 m long (depth) mechanical structure to accommodate the beam transport line. A desirable structurally determined support on two bearings generates large bending moments. To avoid excessive flexure the structure would reasonably be some kind of heavy cylindrical ("barrel type") or conical shell or space-truss, also capable of transferring large torsion forces[19].

A critical part would be the design of the front ring, which will carry most of the load induced by the magnet and the counterweight; its diameter has to allow access for the patient and the PPS, thus restricting propping of the ring to perimeter regions. Forces will have to be carried by bending, resulting in large deformations or very heavy dimensions of the ring (to gain the necessary stiffness). A promising solution to this problem could be to cantilever out the last heavy dipole from the bearing ring, as it is shown at the PSI-gantry.

ECCENTRIC GANTRY SOLUTIONS – "RIESENRAD-GANTRIES"

A possibility to reduce the radius of the gantry dramatically (nearly by the factor of two), is to have the **beam delivery system and the patient eccentric**, as it was done at the proton gantry at PSI. Compared to a conventional gantry the structural complexity will rise. Therefore, this solution shows its power when space is very restricted.

A remarkable simplification of the beam delivery system is achieved by gantries, which deliver the beam from a centrally mounted rotating 90° bending magnet outward onto the eccentrically positioned patient. In imitation of the Vienna Riesenrad such solutions are frequently referred to as "**Riesenrad-Gantries**". It is obvious that such an approach has a higher structural efficiency than any isocentric solution.

Certainly, the patient cabin will be designed as a closed room. The patient will not be aware of his actual position in the gantry room.

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[18] For a summary on existing proton gantries see reference [14].

[19] Compare the proposals for the beam transport in an isocentric ion gantry in reference [15].

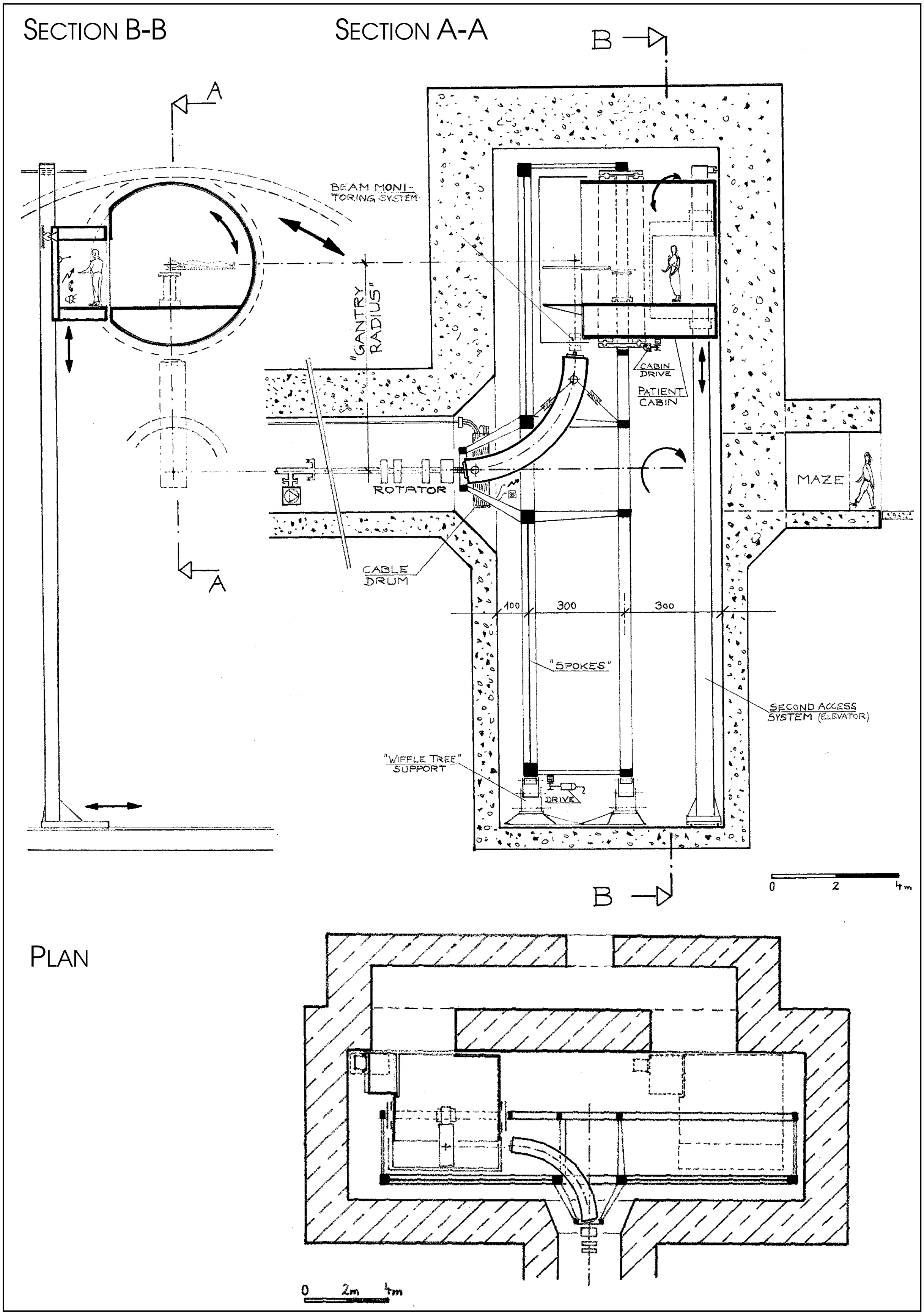

Figure 8: Basic system of a Riesenrad-Gantry, type wheel-gantry. The gantry radius R was taken as 6,6 m. Assumed radius of the bending magnet: 3,5 m.

One possible version of such a Riesenrad-gantry would be to mount the central magnet close to the hub of a **rigid wheel**, which is supported on its perimeter by roller-bearings (figure 8). The overall radius of the machine is slightly larger than the one for a corresponding isocentric gantry because there has to be the possibility to position the patient perpendicular to the original beam axis with his head or toe in the isocentre. Depending on the magnet and scanning system chosen, the wheel must have a radius between 8 m and 10 m.

The access level is placed at medium height and two entrances to the maze are provided. Therefore the maximum rotation of the gantry necessary in an emergency is limited to 90 degrees, corresponding to 15 s when a speed of 1 rpm is assumed.

By cantilevering the cabin from the second bearing wheel, there is space available for a *second access system* which could be a standard industrial *rack and pinion elevator*. The cabin is served from the side, direct access to the maze is via the front. The supporting rails are either mounted on the floor or on the wall. To increase security it is the possible to add another elevator opposite the first one.

The overall depth of the gantry room turns out to be less than 7 m.

The *main bearing ring* of the wheel structure could have pre-stressed spokes, giving the required high rigidity and thus limiting deflections to a reasonable value at a low weight. Basically, the load of the heavy 90° magnet will be carried by normal forces only, making this solution – from the structural point of view – highly efficient. The second ring, weakened by the “hole” for the patient cabin, stabilises the wheel, carries the nozzle (equipped with the beam monitoring system and possibly the scanning magnets) and supports the patient cabin.

CABIN-GANTRY

Another structural approach for a Riesenrad-gantry would be to support the 90°-bending magnet and the eccentric patient cabin by a central axis. For a proton gantry a group from the Harvard Cyclotron developed such a solution during the 80’ies. P. Negri recently presented a preliminary design for a “mobile **cabin gantry”** for ions following the same structural principle [11].

The patient cabin and the opposite counterweight will be mounted on two girders cantilevering from a central axis. The majority of the weight of the 90° bending magnet will be carried directly by the axial space frame. Trusses running between the cantilevering girders stabilise the magnet, carry the nozzle (equipped with the beam monitoring system and possibly two scanning magnets) and stiffen the whole structure. The principle of supporting the patient cabin on its end panels requires the patient cabin to be structural and stiff between its two supports (i.e. the girders). Due to the necessary beam-entrance channel, no full 360° rotation of the cabin is possible. Nevertheless, a 180° rotation of the gantry starting from top position will be sufficient, if the PPS can turn at least 180° around the (local) isocentre (figure 9).[20]

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[20] To allow for irradiation with multiple fields without frequent re-positioning of the patient, the actual PPS rotation possible should be considerably higher than 180°.

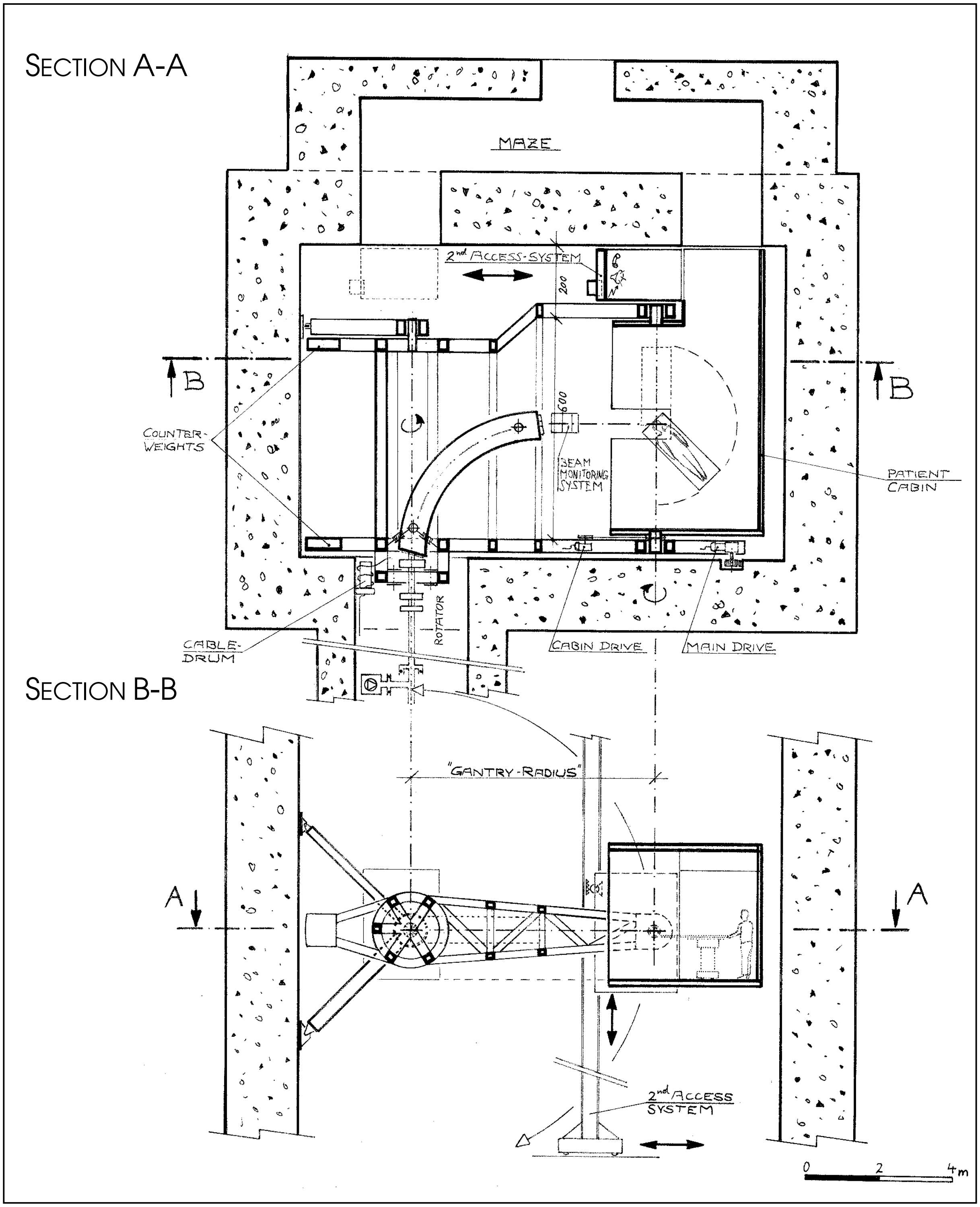

Figure 9: Principal system of a Riesenrad-Gantry, type cabin-gantry. The gantry radius R was taken as 6,6 m. Assumed radius of the bending magnet: 3,5 m.

Eventually the "restriction" turns into an advantage, because the building volume could be reduced by approximately one third (the depth of the gantry room being about 9 m). Unfortunately, the counterweight will become fairly heavy due to the small lever arm.

An alternative would be to support the cabin only by one girder; the cabin would cantilever into the beam line in a similar way as it is shown for the wheel-gantry gaining about 0,5 m in depth. Transition to a 360° rotation would reduce the required space for the PPS, getting another 1 m in depth.

The closeness of the axial bearing and the point of beam delivery from the rotator to the gantry avoids differential deformations at that critical point. Additionally, the structural principle guarantees a virtually fixed distance between incoming beam axis and the (local) isocentre. Unfortunately, the central axis is very heavily loaded and – just like the girders – needs to have large dimensions to avoid excessive flexure.

Inside the gantry room the axis is mounted on an “A”-shaped support, which is fixed to the sidewall. Therefore, the total length of the central axis is kept low and space is made available for the installation of a second access-system, which in principle could be the same as explained above. Its central location with a separate entrance from the maze avoids any interference with the gantry in an emergency situation.

6.6 Building

SHIELDING

Carbon ions used in therapy are more energetic than protons (400 MeV per nucleon and 230 MeV respectively), producing more secondary neutron radiation when interacting with magnets, beam delivery devices and human tissue. This neutron radiation has to be shielded, which can reasonably be achieved by soil, concrete and water.[21]

On the other hand, more dose per nucleon is deposited by ions, which reduces the number of particles necessary for therapy. Due to active scanning, no beam losses at collimators and other devices will occur and most of the produced ions will eventually reach the target tissue. Therefore – as a first approximation – similar or slightly increased shielding requirements as for the proton facilities are assumed, i.e. shielding walls made out of ordinary concrete, approximately 2,5 m thick.

Apart from the different volume, the principal structural gantry solutions show virtually no impact on the shielding requirements.

The secondary neutron radiation spreads out mainly in beam direction. Consequently, it could be advantageous to mount or integrate part of the required shielding locally on the gantry. In particular with isocentric gantry solutions the counterweight could be used for this purpose.

RELATION GANTY RADIUS TO COST

Figure 10 gives a rough impression how different gantry systems and their various radii affect the costs of civil engineering works for the gantry room.

Depending on the soil mechanics situation additional costs for excavation, ground support, ground water control and the foundations might arise, although generous assumptions were already made.

The absolute figures represent the net cost of civil engineering work for a stand-alone gantry room, where walls represent between 40 % and 50 % of the costs. Therefore, considerable cost savings can be achieved when the lateral shielding walls will be shared with other treatment rooms in a hadrontherapy centre. Savings encountered by sharing parts of the rear wall with the switchyard will compensate the additional costs for the maze.

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[21] The photon radiation, which is also produced during irradiation, plays only a minor role [16].

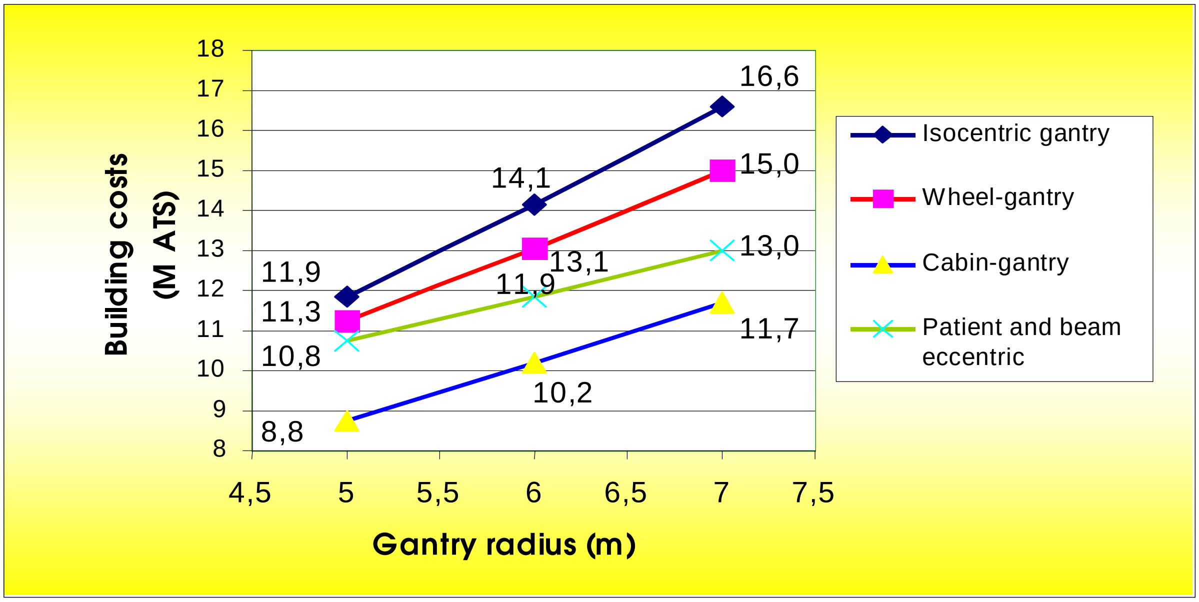

Figure 10: Preliminary cost estimation of the gantry building in dependence of the "gantry radius" comparing several different mechanical gantry solutions, in Mill. ATS (7 ATS = 1 DM). The "gantry radius" always corresponds to the distance between beam axis before the last 90° bending and the target centre. Certainly the actual overall machine radius used for calculation takes into account the enlargement due to the magnet or the patient cabin dimensions. The sums take into account costs due to diaphragm walls, excavation, concrete structure, roofing and façade (no real estate and technical installation costs were considered). The following assumptions were made: half of the building is underground (the beam comes in at ground floor level); no additional area in front of the machine is foreseen; cubic room geometry; foundation slab and walls adjacent to the earth (structural) 1 m other walls and ceiling 2,5 m; added height to accommodate space for the support structure of the bearings 1,5 m (cabin type only 0,5 m).

Due to a possible 30 % reduction of the breadth, space for a cabin-gantry solution is about 2,5 to 3,5 MATS cheaper than the room for a wheel solution (depending on the radius). The latter is again between 0,5 and 1,5 MATS cheaper compared to the isocentric solution. The considerable length (depth) of the room for an isocentric gantry (~17 m) makes this solution – from the "gantry-room-cost" point of view – unattractive.

A meter increase in radius would count for additional costs for the isocentric, the wheel- and the cabin-gantry of about 3 MATS, 2,3 MATS, and 1,8 MATS respectively.

For comparison purposes the building costs of a hypothetical eccentric (patient and beam axis eccentric) ion gantry is also shown in the figure 10 (the actual radius of the gantry was calculated half the value of the effective radius). Although – at a first glance – this approach appears to be the most compact gantry solution, it turns out that the room for a cabin gantry needs considerably less volume and is cheaper.

7 Cost Considerations

So far, feasibility studies presented in this report are not detailed enough to provide reasonable cost estimations for the beam optics (magnets), the various structural gantry solutions and the technical installation in the building (cooling and ventilation).

D. Böhne from the "Gesellschaft für Schwerionenforschung" (GSI) compared costs of various beam delivery concepts in an ion therapy facility [17]. Beam optics and the mechanical structure of an ion gantry are estimated to cost 56 MATS and 6 MATS respectively, making 62 MATS net (8,9 MDM) in total. However, the proposed ion gantry (light weight, high power magnet) would be responsible for considerable parts of the building costs, in particular water cooling and ventilation, which are estimated to sum up to 22 MATS for a facility having one ion gantry, one vertical and one horizontal beam.

According to [18] the price for the *proton* gantries installed at the Northeast Proton Therapy Center (NPTC) in Boston was between 55 and 60 MATS (4,5 M $) pro rata neglecting costs for the building and the control system. If one now assumes a slightly increased price for the case of an ion gantry (say 65 MATS) and takes into account civil engineering costs for the gantry room of about 15 MATS, the total net cost for an ion gantry would be around 80 MATS *plus* proportionate costs for the technical installation and the control system.

Böhne's estimation shows the cost saving potential that lies in the beam transport scheme and the corresponding beam optics, compared to other subsystems. This is a strong argument in favour of Riesenrad-gantries, which perform only the minimum 90° of bending, saving pro rata costs of dipoles, power supplies, vacuum chambers and cooling devices.

8 CONCLUSION AND OUTLOOK

CONCLUSION

Based on sound studies of potential procedures, activities and flows, specifications and requirements for the ion gantry facility and the actual gantry were set. The strategic objectives emerged to be quality, efficiency, flexibility and – crucial to a machine running in an hospital environment – safety. It has to be assumed that the design will only be accepted, if the safety standards are the same as in conventional radiotherapy using isocentric gantries.

Beam-optical considerations show adverse implications in case of (complete) upstream scanning, either for the cooling and ventilation of the gantry room or the gantry structure. This depends on whether a lightweight, high power magnet is used, or a conventional magnet, which would become fairly heavy. Performing one dimension of scanning after the final bending magnet seems to be an efficient solution but the design of the final scanning magnet remains a big challenge to be addressed.

From the structural point of view so-called Riesenrad-gantries (wheel-gantry and cabin-gantry) represent a promising approach to get a competitive ion gantry. Building costs (without technical installation) are proportional to the gantry radius. However, the impact of an increased radius on the overall costs is marginal (a few percent only). This suggests that minimisation of the radius can be a misleading objective, especially when regarding its implications on scanning procedures and magnet costs.

OUTLOOK

The definition of the future ion gantry facility and its processes revealed the basic interrelationships and mutual impacts between the crucial *systems* of an ion gantry. Feasible solutions were generated concerning the issues of access, beam optics, the patient cabin, the gantry structure and the building.

Having the presented program of the ion gantry facility in mind, it shall now be possible to work out and evaluate the most effective general solutions. The next step will see the simultaneous development of the principle structural concepts for the gantry (including cost approximations) and the basic beam optical design. A review, where the program is revised, adapted to new developments and accepted by all disciplines involved, should follow.

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