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STATUS OF THE HZB CYCLOTRON

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Abstract

For more than 20 years eye tumours are treated in collaboration between the Helmholtz-Zentrum Berlin (HZB) the Charité – Universitätsmedizin Berlin (Charité).

The close co-operation between Charité and HZB permits joint interdisciplinary research. Irradiations with either a sharp, well focused or a broad beam, either in vacuum or in air are possible. In the past few years, we concentrated on beam delivery for FLASH experiments and the related dosimetry. For example, intraocular lenses have been irradiated under normal and FLASH conditions to investigate possible changes in the translucence. Furthermore, radiation hardness tests on solar cells for space have been performed.

A modernization project has been started in order to secure a long term and sustainable operation of our accelerator complex for therapy and research.

The accelerator operation for therapy as well as ongoing experiments and results will be presented.

ACCELERATOR OPERATION

A layout of the facility can be found in Ref. [1]. Either the 2 MV Tandetron or the 6 MV Van-de-Graaff serves as injector into the k=130 cyclotron of HZB. Overall, accelerator operation went quite well in the past years (see Fig. 1). Only in 2015 and in 2021 the downtime was above 5%.

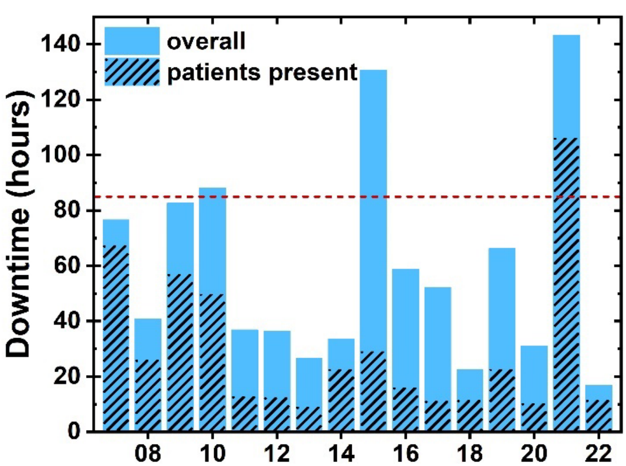

Figure 1: Downtime in hours for the past years. With exception of 2015 and 2021, the relative downtime was below 5%.

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In 2021, a water leak in the extraction magnet required an opening of the cyclotron. For this reason, the scheduled therapy week had to be suspended. This explains the large duration of downtime with patients present, as the postponement is counted as downtime.

Fortunately, a new coil for the extraction according to the original plans had already been acquired. Hence, a rapid 1:1 exchange was possible. In the extraction magnet resides also a correction coil (see Fig. 2), which has never been used in the past. After the exchange of the coil, we failed to extract the beam. Only after connecting and using the correction coil the beam could be extracted with the usual transmission. As the magnet has adjust pins and the extraction coil fits snugly in the magnet, it is most probably the vacuum chamber which is now on a slightly different position.

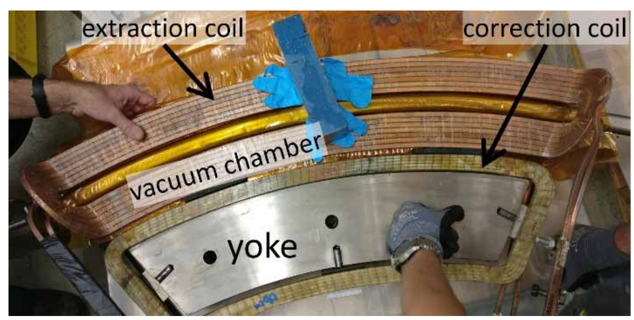

Figure 2: The extraction magnet with extraction and correction coil. On the yoke, the adjust pins are visible.

BEAM UTILIZATION

Roughly 83% of the beam time is used for therapy, the other 17% for experiments. Experiments comprise accelerator research and development (9%) [2, 3] as well as medical physics and dosimetry, (4%) e.g. [4, 5]. The distinction between the two topics is sometimes difficult. The final 4% of the beamtime is used for radiation hardness testing by external users, e.g. [6, 7].

Beam Delivery for FLASH Irradiations

The definition of FLASH irradiation is the delivery of the dose in a time of less than 1 s with a dose rate above 40 Gy/s [8]. The idea is to maintain tumour control and minimize side effects. The high dose rates pose challenges both for dosimetry in terms of linearity and saturation effects, and for reliable beam delivery in terms of beam stability and providing the same dose from shot to shot.

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The short irradiation times led to a closer look on the switching times: The Faraday cups with a stroke of 50 mm are too slow for this purpose, as they operate within 0.1 s. The special mechanical, scissor-like beam shutter used for therapy (BSATT) is faster, with an opening time of ~ 10 ms and a closing time of ~ 5 ms. This is more than sufficient for our conventional irradiation times of 30 s to 60 s. However, in order to provide a homogenous field this limited the achievable FLASH times to 200 ms.

Furthermore, the delays between beam demand and full intensity as well as stop command and zero beam have to be considered (see Fig. 3). The real irradiation starts with a delay time of t_1, then there is the time t_2 while the shutter opens, after t_3 the request to stop the beam occurs, however, there is the delay time t_4 until the shutter starts moving and closes within time t_5. The area under the curve corresponds to the applied dose. The dose applied within time t_4 and t_5 is called excess dose, because it arrives after the stop signal has been triggered.

In order to provide a dose controlled irradiation instead of setting a fixed time, a LabVIEW code and a FPGA was programmed. The FPGA board processes the signals of the Faraday cup and two ionization chambers. The FPGA board controls the Faraday Cup, the fast beam stopper and the electric beam deflector.

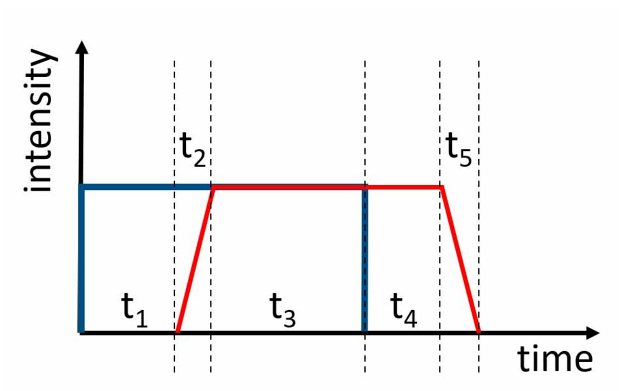

Figure 3: the blue line depicts the ideal FLASH irradiation, the red curve the reality. The different times to be considered are the delay between beam request (t=0) and start of opening of the beam shutters (t_1), opening time (t_2), time between request to interrupt the beam until (t_3) start of closing (t_4), closing of the shutter (t_5).

In the so-called calibration mode, the irradiation time is fixed, and the dose is measured via a Markus chamber (PTW Freiberg) and entered manually. The code correlates the sum of the counts of the ionisation chambers to the measured dose and integrates the counts after the stop-signal was triggered (excess counts). This defines the switch-off strategy: in "dose mode" the stop-signal is given for the corresponding dose minus expected excess counts.

After the calibration several trial runs are performed to verify the set-up for the day and to determine the dose fluctuation from shot to shot. This takes only a few minutes. In the conventional irradiation mode applying 15 Gy with 0.25 Gy/s and 60 s total irradiation time, statistically no fluctuations were observed. In FLASH mode with a dose rate of 75 Gy/s after 30 trial runs a mean dose of 14.9 Gy with a standard deviation of 0.6% was determined. Thus, since then FLASH irradiations with a very high dose accuracy are now possible.

Irradiation of Mice Eyes under Conventional and FLASH Conditions

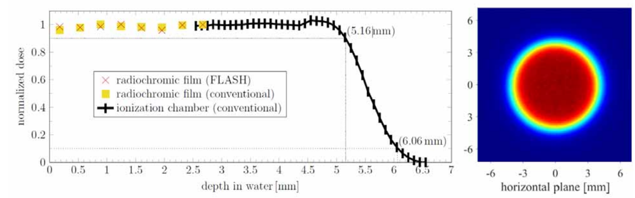

Figure 4: The field size for the irradiation of single mouse eyes. Left: Spread out Bragg Peak with a range of 5.2 mm and a distal fall-off of 0.9 mm, measured with ionisation chamber and radiochromic film. Right: Lateral beam distribution measured with a camera, showing a field size of 6.3 mm and a penumbra of 1.7 mm.

Mice eyes are small, thus providing challenges for the beam delivery. The FLASH irradiations for mice eyes should provide the same irradiation field as in the conventional irradiation in order to irradiate a **single** eye, thus leaving the other one for comparison purposes: a range of 5.2 mm in water with a 0.9 mm distal fall-off as well as a lateral field size of 6.3 mm with a lateral penumbra of 1.7 mm (see Fig. 4). Distal fall-off and penumbra are both from 90% to 10% of the isodose. This was achieved by aluminium for beam scattering and range shifting and a brass collimator. The desired full spread-out Bragg Peak was achieved by a modulator wheel providing 960 spread-out Bragg Peaks per second [9]. Figure 5 shows the experimental set-up.

FLASH irradiations with an extended Bragg Peak and a good confinement to the target volume are now feasible, the final data evaluation is still on-going [10].

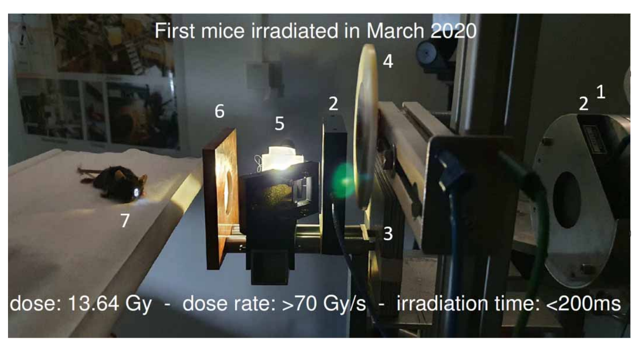

Figure 5: The set-up for the Flash irradiation of single mouse eyes: 1 vacuum window, 2 monitor chambers (7861, PTW-Freiburg, Germany), 3 combined degrader and scattering system (16 mm Al), 4 3D printed modulator wheel (periods: 48; rotation speed: 20 Hz), 5 light field system (LED lamp and mirror foil (25 µm Kapton), 6 collimator (brass; Ø5.5 mm), 7 anesthetized mouse with light field for positioning.

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Further Developments of FLASH Irradiations

At the moment the best irradiation time and the best dose rate are unknown. There is some evidence, that the times should be shorter than 100 ms [11]. Then the mechanical shutter is too slow and has to be replaced by an electrical deflector. Until recently, only for the Van-de-Graaff injector such a deflector was available, not for the Tandetron, the standard injector for proton therapy. Such a deflector has been developed and is now available [12]. Furthermore, a transmission ionisation chamber with a linear dose response up to 400 Gy/s has been developed [13].

The next steps of further development of the experimental set-up are the replacement of the low-efficient single scattering beam shaping system by a double scattering system and the replacement of a rotating modulator wheel by ridge filters. In the near future, FLASH irradiations of sarcoma and uveal melanoma organoids will be performed.

Determination for Appropriate Material for Intraocular Lenses in Ocular Proton Therapy

Intraocular lenses from various manufacturers have been irradiated with protons under conventional and FLASH mode in order to investigate changes in the material. The lenses were mounted in the water phantom just behind the entrance window with a Markus chamber for absolute dosimetry (see Fig. 6). The irradiation field size was identical for both modes. In conventional mode 60 Gy were applied with a dose rate of 0.2 Gy/s, while in FLASH mode the same dose was applied with a dose rate of 70 Gy/s.

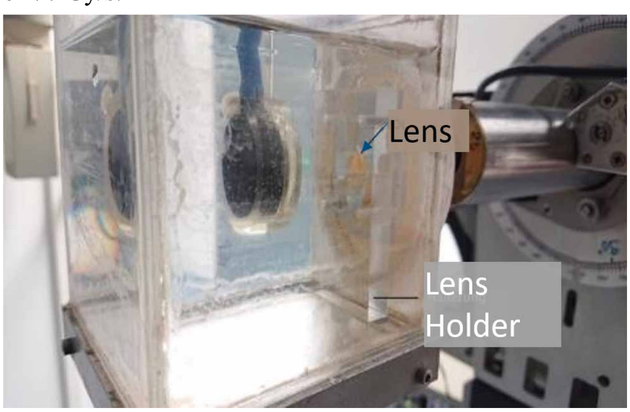

Figure 6: The lens is mounted inside the water phantom right behind the entrance window with a Markus chamber behind for dosimetry. The proton beam is coming from the right.

Lenses from different manufacturers with different dioptres (D) have been irradiated:

- Bausch+Lomb “enVista” +10.0 D Modell MX60 with UV absorber
- Bausch+Lomb “Akreos” +11.5 D with UV absorber
- Bausch+Lomb “Softport” +17 D
- Zeiss “CT Lucia 201P” + 19.5
- Zeiss “CT Asphina 509 MP” +21 D

Before and after the irradiation the transmission of the lenses was measured for light ranging from 200 nm to 900 nm. One example is given in Fig. 7. Only slight, but non-relevant changes in the transmission before and after the irradiation could be observed, the edges and peaks in transmission were at the same wavelength. The result was the same for conventional and FLASH irradiation, meaning that the dose rate is irrelevant. Only after an irradiation with ^{60}Co γ-rays with a dose 20000 Gy, a typical sterilisation dose, changes in transmission were observed.

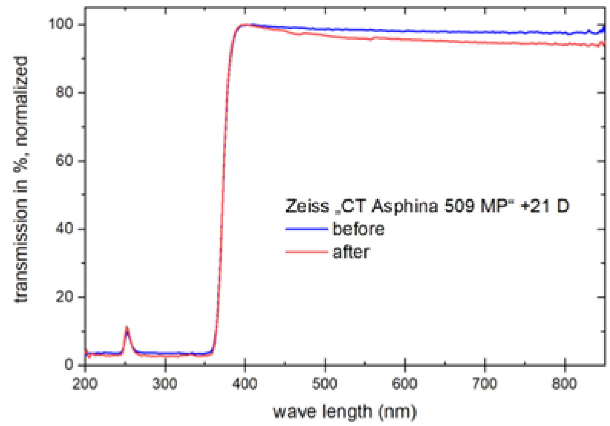

Figure 7: Normalised transmission curves for the Zeiss CT Asphina 509 MP +21D lens before and after the irradiation. Only slight changes are visible.

CONCLUSION

Over the past years the cyclotron operation went quite smooth, with just one major event leading to the postponement of a therapy week. Having the spare part, the extraction coil, already in house, helped tremendously. In total, more than 4400 patients have been treated and, in spite of Covid-19, patient numbers were high.

Beam delivery for FLASH irradiations has been developed, allowing now dose controlled FLASH irradiations. For this purpose, the different times playing a role in the process have been investigated and are now handled via a FPGA board together with a LabVIEW code.

We have developed an irradiation set-up which allows the irradiation of single mouse eyes, thus leaving the other eye for comparison. This set-up operates under conventional as well as under FLASH conditions.

Intraocular lenses have been irradiated under normal and FLASH conditions. No significant changes in transmission could be observed.

In near future irradiations of sarcoma and uveal melanoma organoids are scheduled.

Furthermore, experiments for dosimetry and improved beam delivery, especially for FLASH conditions will be performed as well as irradiations of electronic devices and solar cells for radiation hardness tests.

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