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15th Triennial World Congress, Barcelona, Spain

TRAJECTORY CONTROL FOR A NEW TOTAL SKIN ELECTRON TREATMENT MACHINE

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Abstract: The paper focuses attention on the development of a control system for a new radiotherapy treatment machine that makes use of a betatron to deliver electron beams over the whole surface of a patient. The betatron moves above a patient at a fixed distance from the patient support system. This paper presents a method to predict the betatron trajectory, in terms of speed and position, according to variation in the patient's geometry and is concerned with a simulation study of a PID control strategy implemented to control the speed of the betatron. *Copyright © 2002 IFAC*

Keywords: conventional control, medical systems, medical applications, three term control, optimisation.

1. INTRODUCTION

Total skin electron (TSE) treatment is an effective procedure to treat skin cancer such as Mycosis Fungoides (Klevenhagen, 1985; Shank, 1998). TSE makes use of electron beams to treat the skin over the whole surface of the patient whilst, at the same time, sparing tissues underneath the skin. Indeed, electron beams interacting with human tissues only penetrate 10 to 15 mm before being significantly attenuated. At the University Hospitals Coventry and Warwickshire NHS Trust, Coventry, UK, TSE is currently delivered using dual energy linear accelerators. Linear accelerators are machines capable of producing X-ray photon or electron beams. The current treatment technique is, however, time and resource consuming, requiring patient specific immobilisation devices to be manufactured and used to set-up the patients. In addition, TSE takes longer to administer than standard radiotherapy treatments. This means that each patient treated for TSE on linear accelerators can potentially increase the number of patients waiting for the more routine radiotherapy treatment. Given the significant pressures imposed on the NHS to reduce waiting lists, this problem needs to be addressed. It was therefore decided to build a specialised and dedicated TSE treatment machine capable of irradiating the whole body with an electron beam more efficiently and cost effectively than using linear accelerators.

This paper introduces the conception and design of the new TSE treatment machine, referred in the remainder of the paper as the TSE unit, and focuses on a method to predict and control its trajectory, in terms of speed and position. Section 2 presents the background to the work and the design of the TSE unit. Section 3 describes the electron beam model developed to predict the penetration of the electron beam depending on the distance from the source, the angle of incidence of radiation and the dose rate. Section 3 also presents a derivation of the expression used to predict the speed of the trolley supporting the betatron. Section 4 presents a study on the effect of the electron beam field size and dose uniformity on the treatment time. The last section presents the control approach and optimal tuning of a three term proportional + integral + derivative (PID) controller.

2. DESIGN OF THE TSE UNIT

A number of elements were considered during the design of the TSE unit. These include the method of generating electron beams, the means to deliver a uniform dose over the whole body of the patient and the cost effectiveness of the new machine.

The first element selected was a machine, namely a betatron, capable of generating an electron beam at the appropriate power. A betatron is a cylindrical electron accelerator that was first used in medical applications in 1948 (Klevenhagen, 1985), Figure 1. The betatron's weight and small size together with the ability to produce 6 Mega electron volts (MeV) electron beams at a fraction of the cost of a linear accelerator, makes the betatron an ideal tool for TSE.

Having selected the means to deliver an electron beam of sufficient power, the second element was to design a system that could irradiate the whole body of the patient. The emphasis was placed on robustness, safety and comfort. Following a call to enter into an open competition, organised by the CTAC, Coventry University, UK, (Hall, 1998), two alternative designs emerged. The first design, that offered the most flexibility, involved attaching the betatron to a six degrees of freedom industrial robot. However, due to the potential risk of collision with the patient and the additional stress put on the betatron, it was decided to implement a solution based on a fixed gantry construction, see Figures 1 and 2. As opposed to a solution published in (Chretien *et al*., 2000) where a variable speed translation couch was developed to move the patient under a fixed treatment machine, it was decided that the patient support system would remain fixed. The betatron, positioned onto a trolley 140 cm above the treatment couch, is allowed to move along the patient's head to toe axis (denoted '*x*' axis) (Guerin, 1999; Haas *et al*., 1999), see Figures 1 and 2. Such a solution was adopted to minimise the patients discomfort during treatment. The authors believe that this should ensure better reproducibility, minimise patients movement and reduce patient stress.

Other aspects to consider when developing a treatment machine are aesthetic and comfort factors. The betatron is noisy. To attenuate the decibel level a special casing with noise absorbing material was constructed. A false wall was also fitted in front of the gantry and betatron so that the patient would only see the electron beam aperture.

Finally, the most relevant aspect to a control engineer is the need to design and develop a control system to control the displacement of the trolley supporting the betatron. To drive the trolley, the SA28 (Smartdrive Ltd 2002), a complete stepper motor system, was purchased. It includes the controller Euro205 that was chosen for its expansion capabilities, allowing sophisticated controllers to be implemented.

TSE unit during construction

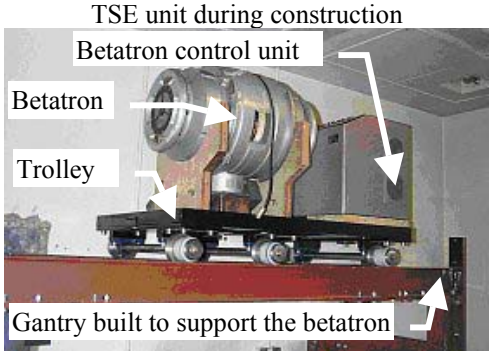

Fig. 1. Illustrating the betatron and beam control system mounted on a moveable trolley positioned on rails on a purpose built gantry.

Set-up used to treat patients using a betatron

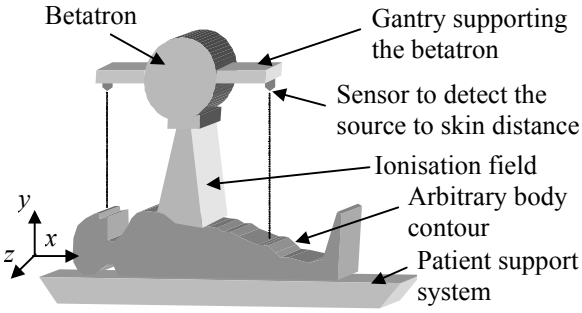

Fig. 2. Schematic representation of the set-up used to move the electron beam produced by the betatron over the whole length of the patient.

The trolley (1.1 m long by 0.75 m wide) is positioned onto rails (3.5 m long and 0.75 m wide) and its four nylon wheels (0.075 m diameter) are designed to minimise the amount of friction with the rail. This means that a relatively small force needs to be applied to move the trolley. However, care should be taken as the moving system composed of the trolley, the betatron and the betatron control unit weights in the order of 300 kg. This is achieved by constraining the control input to maintain the momentum of the machine within acceptable bounds. Such an arrangement should prevent unnecessary strain on the motor gearbox and transmission as well as within the drive belt.

It can be observed from Figure 1 that the height of the betatron cannot be modified. The only means of changing the distance from the beam to the patient is by modifying the height of the patient support system. To minimise the movement of the patient during treatment, the position of the couch is adjusted once before the commencement of the treatment and kept fixed whilst the betatron mounted onto the trolley irradiates the patient. Therefore, in this work, the distance from the couch that supports the patient to the betatron is assumed constant. The energy reaching the patient depends on the distance between the source of the electron beam and the surface of the patient. Naturally, this distance between the source and the skin will vary depending on which part of the body is being irradiated and the geometry of the body

contour in the (*y, z*) plane. To deliver a constant dose, it is therefore necessary to modify the output of the betatron, referred to as dose rate, and/or to vary the time of exposition to the electron beam.

Previous work in the area of beam intensity modulation has demonstrated that modulating beam intensity in real time is not a trivial problem. It requires state of the art radiotherapy machines, such as linear accelerators used with multileaf collimators or patient specific compensators (Haas, 1999). Whilst the latter is applicable to standard radiotherapy treatment using X-rays, the need to adapt the beam intensity profile to the patient geometry, along the *y* axis, across the full length of the patient makes it impractical for TSE. Further, the betatron is designed for simplicity to ensure 'better' robustness than complex radiotherapy machines. Such robust design does not permit modification of the beam output in real-time. Inserting compensators in the beam is not, in this case, a realistic option as it would attenuate the electron beam too significantly. Therefore, the solution adopted in this work, to ensure that the patient receives a uniform dose over the whole body, is to vary the speed of the trolley for a fixed field size. When the speed is low the time of exposition to radiation is large and the dose delivered to the patient is correspondingly high. As the speed of the trolley increases, the exposure time decreases together with the dose received by the patient. In the future, it should be possible to move the betatron at various angles along the (*x, z*) plane, however the current TSE unit has a fixed beam position. The orientation of the electron beam (30 degree from the *z* axis) has been calculated to offer the best compromise between the area covered by the radiation and the accuracy of the dose delivered. Using such a set up, the treatment is delivered in four passes. During the first pass, the betatron irradiates the front right hand-side of the patient. During the second pass, the couch is rotated by 180 degrees and the left hand side of the patient is then irradiated. Then the patient is turned over and again the right and left hand-side are irradiated.

3. SYSTEM MODELLING

To facilitate the understanding of the challenges to be solved both in terms of modelling and control, it is assumed, in this section, that the electron beam is directly above the patient in the *y* axis direction. It is also assumed that the beam can be decomposed into a number of elemental pencil beams i.e. i^{th} pencil beam represented by a white dashed line in Figure 3.

3.1 Electron beam modelling

Factors considered when modelling the effect of electron beams on human body structures include field size, dose rate, distance of the patient from the source of radiation and angle of incidence of the beam.

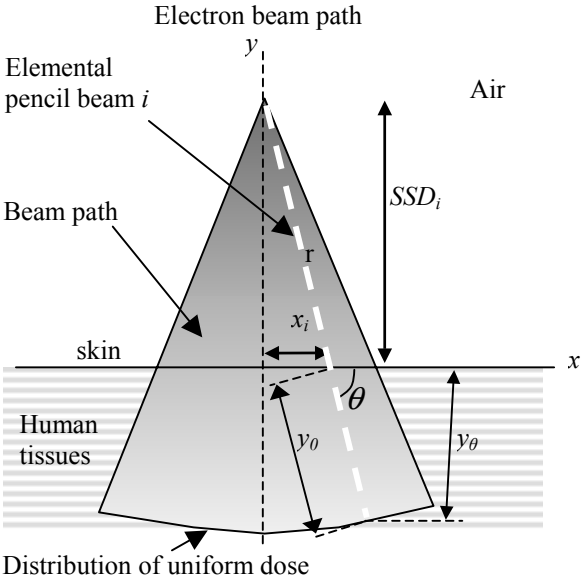

Fig. 3. Illustrating the concept of pencil beams and the spherical distribution of the beam's energy into an assumed flat portion of skin.

The field size is defined as the size of the field exposed to radiation. As the field size increases, the energy lost due to multiple scattering between the electrons and the air decreases. This, leads to an increase in the energy delivered to the patient. To account for this effect, the original energy is multiplied by a so-called output factor; denoted OF. In this work, the field size can be adjusted from 6 cm^2 to 40 cm^2 corresponding to output factors of 1 and 1.043 respectively (Guerin, 1999).

The dose rate is the quantity of energy delivered per unit of time. In this work, it is assumed that the betatron delivers 1.4 Gy/min. The dose rate, denoted I_{ref}, is normalised and given at a distance of 1 meter from the source. The distance from the source of radiation to the patient is referred to as source to skin distance, denoted SSD. In this work, the SSD has been selected to be 140 cm to offer the best compromise between sufficiently large field size and dose rate at the patient surface. The dose rate *I* for a field size associated with an output factor OF at a distance $r=SSD$ from the source is given by

$$I = OF \times I_{ref} \Big/ r^{2.5} \qquad (1)$$

The angle of incidence of the beam is the angle between each pencil beam and the normal to the patient. In this work, consideration is given to the patient head to toe axis (or longitudinal direction), therefore, the angle of incidence in the lateral direction is not considered. The angle of incidence in the longitudinal direction, denoted θ is taken between the line made by the pencil beam central axis and the slope of the patient surface in the longitudinal direction, see Figure 3. The angle of incidence influences the penetration of the electron beam, with the highest penetration achieved for an angle normal to the patient's surface, i.e. $\theta = 90^{o}$.

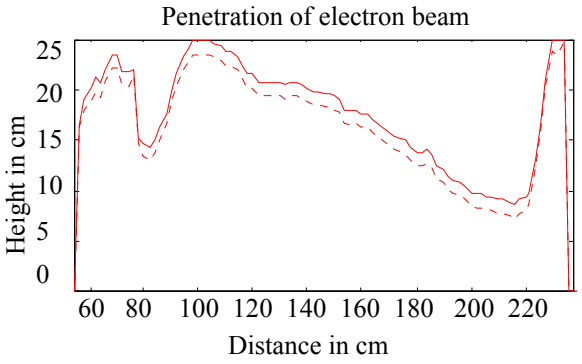

Fig. 4. Illustrating the beam penetration with respect to the patient geometry, patient contour '—', dose penetration '- -'.

The penetration is given by the following expression:

$$y_\theta = y_0 \left|\sin(\theta)\right| \tag{2}$$

where y_0 and y_θ are, respectively, the penetration with an angle of incidence normal to the patient and a beam incident at an angle θ from the horizontal axis, see Figure 3. An arbitrary profile given in Figure 4 shows that for most of the body the penetration of the beam is adequate with the exception of the area under the chin, the top of the chest and the feet.

To improve the dose uniformity, it has been suggested by (Guerin 1999) that i) patients could stretch their feet and ii) the patient support system could be tilted resulting in a patient surface orthogonal to the beam central axis. However, as the latter two solutions involve moving the patients, it was proposed that the betatron should be mounted on a gantry that can be tilted such that the electron beam can be orthogonal to the patients surfaces.

3.2 Determining the speed of the betatron

Radiotherapy treatments such as TSE are in general delivered in a number of fractions, each fraction accounting for a part of the overall dose prescribed. In this work, it is assumed that a dose of 24 Gy is delivered in 12 fractions of 2 Gy. A two-step process has been adopted to calculate the trajectory of the beam that is assumed to comprise n elemental pencil beams. The first step is to calculate the dose rate at the skin level at the point of intersection between each elemental pencil beam, denoted i =1...n, and the patient's surface. The second step is to determine the speed of the betatron assuming that the motor follows a continuous movement with a fixed field size. As the patient is non-flat, the dose rate at the surface of the patient changes. Regions such as the neck or the legs are further away from the beam source than the chest or the abdomen and will, therefore, require longer exposition to receive the same amount of dose. In order to take this variation into account, the average dose rate for a given field size, denoted I_{avg}, due to each elemental pencil beam i for the points within the open field is calculated as follows:

$$I_{avg} = \frac{\sum_{i=1}^{n} I_i}{n} = \frac{1}{n}\sum_{i=1}^{n} \frac{OF \times I_{ref}}{\left(\sqrt{SSD_i^2 + x_i^2}\right)^{2.5}} \tag{3}$$

with i=1..n, is the index corresponding to the pencil beams considered, n is the number of pencil beams, OF is the output factor, I_{ref} is the dose rate at 1 m, SSD_i is the source to skin distance along the y axis, and x_i is the distance from the point considered to the beam central axis. As the trolley moves, each point in the patient receives a dose from each pencil beam (that are oriented at different angles due to the beam divergence). The speed of the betatron can therefore be obtained from the average dose rate such as:

$$\dot{x} = \frac{\delta x}{\delta t} = \frac{\Delta x}{\frac{D_p}{I_{avg}}} = \frac{\Delta x \times I_{avg_i}}{D_p} \tag{4}$$

where Δx is the distance along the x axis between two consecutive points on the body, D_p is the prescribed dose and, I_{avg} is the average dose rate.

4. EFFECTS OF SPEED MODULATION AND FIELD SIZE OVER TREATMENT ACCURACY

Having established a model to predict the speed of the betatron from the source to skin distance, this sub-section investigates the effect of speed modulation and field size on the accuracy of the dose delivered to the patient.

Given a required dose per fraction of D_p=2.0 Gy delivered using a 20 cm^2 field size, and knowing the length of the patient and the average distance from the source, it is possible to deduce an average speed for the betatron Unit (here 8.8 cm/min). It can be observed in Figure 5 that such a scheme lead to a dose received by the patient varying between 1.6 Gy and 2.3 Gy. By contrast, when variable speed is used the dose received by the patient is much more uniform, with some small areas receiving dose around 1.8 Gy and 2.2 Gy. The improvement realised by varying the speed is significant, however, the feet together with the head are over-dosed whilst the ankles and the neck under-dosed. Using a field size of 20 × 20 cm means that at any one time an area of 20 cm^2 is irradiated. The speed is calculated from the average distance between the beam source and the region irradiated. If the region irradiated contains body surfaces that change significantly in the y direction, then body surfaces that are closer to the source will receive higher dose than those further away.

In an attempt to further improve the accuracy of the dose delivered to the patient, simulation studies have been performed to determine the influence of field size and the *SSD* to the dose uniformity. Figure 6 illustrates the variation in terms of the dose received

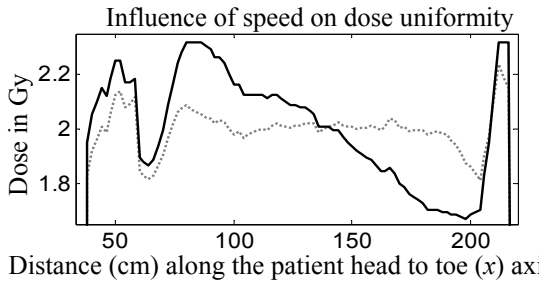

Fig. 5. Illustrating the dose received by a typical patients profile with constant (black solid line) and variable (grey dotted line) TSE unit speed.

by the skin of the patient for a fixed field size and a speed varying according to the moving average of the dose rate at the patient level (dose represented by 'o', corresponding to continuous motion). It can be observed that large field sizes are not able to adapt to the patients profile with the dose distribution being inversely proportional to the SSD. Smaller field sizes are able to produce a more homogeneous dose. The latter improvement is however achieved at the expense of a longer treatment time; it is increased by a factor of more than 3 when a 6 cm^2 field is used as opposed to a 20 cm^2 field, see Table 1. Note that, in Table 1, the numbers in brackets represent the dose distribution obtained without the additional boost.

Table 1: Illustrating the trade off between treatment time and accuracy of dose delivered

Field size (cm^2)		6	20	40
Mean Dose (Gy)	+++	2.01	2.02	2.04
	ooo	(1.99)	(1.99)	(2.00)
Variance	+++	0.014	0.035	0.047
	ooo	(0.019)	(0.051)	(0.074)
Standard deviation	+++	0.025	0.057	0.082
	ooo	(0.035)	(0.076)	(0.11)
Treatment time (min)		110	32	16

The foreseen increase in treatment time for small field sizes makes the latter unpractical therefore, an alternative strategy was investigated. It involved delivering the dose with the largest field available and then adding dose boosts using a step and shoot approach with a smaller field size. The latter could be delivered semi continuously by stopping the betatron and changing the field size. However, as such operation cannot be carried out instantly, the following approach is being considered for implementation. Once the first pass has been preformed with a large field size, the machine field size is reduced to a sufficiently small field and the betatron unit is moved to the required locations where a dose boost is delivered.

The plots represented by '+' in Figure 6 show the improvement made from the additional boost method. It can also be observed that such method is mainly useful when large field are used.

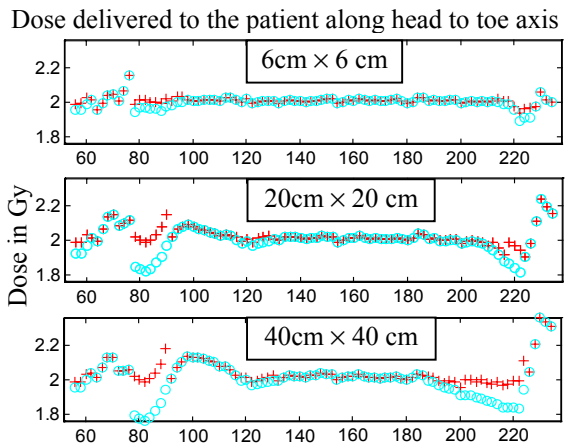

Fig. 6. Illustrating the deposition of dose for fixed field size and variable speed 'o' and with additional boost '+' using a 6 cm^2 field size.

As it is not possible to remove dose, when calculating the initial speed it may be of advantage to select the maximum speed as opposed to the mean speed. This would ensure that no region is overdosed. The latter approach has however the disadvantage of requiring greater dose boosts, which in turn slows the overall process.

5. CONTROL OF THE BETATRON UNIT

The previous sections have highlighted the complexity of the TSE unit control problem in terms of trade off between accuracy of the dose delivered to the patient and the time necessary to deliver such a dose. In this section, it is assumed that the set point (the required speed of the betatron) has been calculated, taking into account relevant clinical requirements together with practical considerations. In the actual system, additional safety constraints are considered to accommodate for possible machine breakdown. The most serious event would be for the betatron to malfunction. When such an event is detected, the power required by the betatron to generate an electron beam is automatically halted and the trolley stopped. To detect such malfunction, the dose rate is monitored constantly. When the dose rate varies within predefined limits, the speed can be adjusted to take into account those variations. However, when such variation becomes unacceptable, the machine is stopped and the treatment interrupted. Such unacceptable variations together with the general health of the machine could be detected using techniques similar to those investigated in (Haas *et. al.,* 2001).

A Simulink™ model of the betatron and its motion has been developed. The control scheme is based on a proportional integral derivative (PID) scheme with a feedforward gain. Such a controller has been selected as it is widely available in industry. A

system model was developed from a number of step responses of the system. A recursive least squares algorithm was used to identify the coefficients of a second order transfer function. The latter was implemented within Simulink™.

An optimisation scheme has been developed in-house to tune controllers according to requirements in terms of: maximum peak overshoot (MPO), settling time (ST), number of oscillations (NO), integral of absolute error (IAE) and the sum of the square of the error (SSE). The optimisation algorithm is based on Nelder-Mead simplex method implemented in the MATLAB™ 6.1 (MATLAB, 2001) function 'fminsearch.m'. In this work, the following objective function was used to optimise the PID gains (Kp, Ki and Kd) together with the feed-forward gain Kf:

$$C = (100*MPO)^2, + (SSE)^2 + (100*ST)^2 + (PEN)^2 \quad (5)$$

where PEN is a penalty function set to an arbitrarily large value. The latter is required as the selected optimisation routine does not constrain its solutions to be positive. The gains obtained for Kp, Ki, Kd and Kf are 0.318, 0.997, 0.193, 0.445, respectively.

The overall system was simulated in Simulink™, with the set-point calculated to produce a uniform dose at 200 cGy using a 6 cm^2 field size, see Figure 6. Figure 7 shows the required and actual speed achieved by the betatron unit. To minimise mechanical stress, the speed of the betatron unit is ramped up to the required speed with the beam turned off. Before passing over the patient, the beam is switched on and once it becomes stable the treatment starts. Once that the whole patient has been irradiated in the head to toe direction, the speed of the betatron unit is ramped down. It can be observed that, as expected, the speed of the betatron follows the same pattern than the patient profile, the smaller the source to skin distance, the faster the betatron is and the smaller the dose delivered.

6. CONCLUSIONS

This paper has presented on-going work in the area of systems modelling and control engineering applied to skin cancer treatment using total skin electron. The design of a new skin cancer treatment unit based around a betatron and a fixed gantry construction was described. A model was developed to simulate the effect of electrons in patients in terms of source to skin distance and angle of incidence of the electron beam. Based on such a model, simulation studies carried out with Matlab/Simulink have highlighted the difficulty in determining a set point for the betatron control unit. It was shown that the set point (in terms of speed of the betatron unit, is a trade off between the accuracy and the time required to deliver the treatment. The simulation studies have also demonstrated the suitability of PID strategy to control the total skin electron unit. The speed calculation algorithm developed to determine the trajectory of the betatron has illustrated the ability to deliver a uniform dose of radiation over a non flat surface.

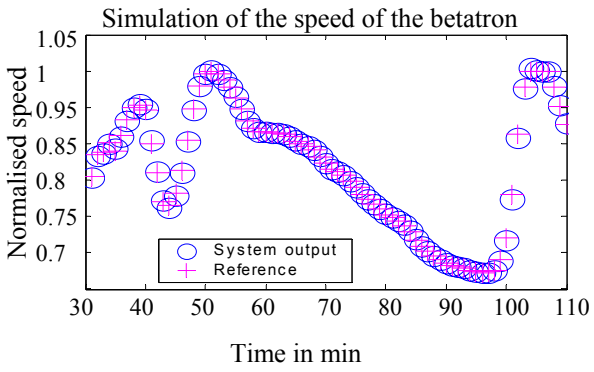

Fig. 7. Comparison between normalised predicted '+' and simulated 'o' speed of the betatron unit to deliver 200cGy using a 6 cm^2 treatment field.

ACKNOWLEDGEMENT

This work was supported by the TRW Foundation.

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