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Technical Notes

The influence of physical wedges on penumbra and in-field dose uniformity in ocular proton beams

Colin Baker [a,*], Andrzej Kacperek [b]

[a] Radiotherapy Physics, The Royal Berkshire NHS Foundation Trust, Reading, UK
[b] National Centre for Eye Proton Therapy, The Clatterbridge Cancer Centre NHS Foundation Trust, Wirral, UK

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ABSTRACT

A physical wedge may be partially introduced into a proton beam when treating ocular tumours in order to improve dose conformity to the distal border of the tumour and spare the optic nerve. Two unwanted effects of this are observed: a predictable broadening of the beam penumbra on the wedged side of the field and, less predictably, an increase in dose within the field along a relatively narrow volume beneath the edge (toe) of the wedge, as a result of small-angle proton scatter. Monte Carlo simulations using MCNPX and direct measurements with radiochromic (GAFCHROMIC® EBT2) film were performed to quantify these effects for aluminium wedges in a 60 MeV proton beam as a function of wedge angle and position of the wedge relative to the patient. For extreme wedge angles (60° in eye tissue) and large wedge-to-patient distances (70 mm in this context), the 90–10% beam penumbra increased from 1.9 mm to 9.1 mm. In-field dose increases from small-angle proton scatter were found to contribute up to 21% additional dose, persisting along almost the full depth of the spread-out-Bragg peak. Profile broadening and in-field dose enhancement are both minimised by placing the wedge as close as possible to the patient. Use of lower atomic number wedge materials such as PMMA reduce the magnitude of both effects as a result of a reduced mean scattering angle per unit energy loss; however, their larger physical size and greater variation in density are undesirable.

Introduction

In ocular proton therapy, physical wedges are used in order to spare a critical tissue, e.g. optic disc or macula, or more usually, to reduce the high dose volume. Approximately half of practising ocular proton therapy centres employ wedges. For example, at the Helmholtz Zentrum Berlin [1] and at The Clatterbridge Cancer Centre (CCC), wedges are used in 70% and 30% of cases respectively. The wedge is usually mounted on the beam collimator by means of a stalk arrangement. It is normal practise to position the wedge as closely as possible to the patient's eye to minimise the effects of proton scatter, but consistent with patient safety. The stalk length may typically vary between 15 and 40 mm depending whether the wedge is planned on the temporal or the nasal side of the treatment eye. On the temporal side of the patient it is possible to approach within 10 mm, but on the nasal side proximity of the wedge is limited by the patient's nose and can be up to 30 mm from the patient surface. Wedges, typically made of aluminium or PMMA, have angles between 10 and 60 degrees, defined in terms of angles in eye tissue (assumed here to be similar in composition to water, with a density of 1.05 g cm^{-3}).

The effect of small-angle proton scatter from the edge (toe) of a wedge partially placed in a proton beam has been noted previously [1,2]. It was described and applied to proton radiography in the early 1970s by West and Sherwood [3]. Goitein et al. [4] later modelled this effect for both electrons and protons incident on thin absorbers, providing guidance for clinical proton therapy. The effect of proton scattering on fluence and consequently on absorbed dose is shown in Fig. 1 for both a plane and a wedged absorber when placed partially into the beam (the geometry of which is indicated in Fig. 2). Disequilibrium in scatter contributions is observed near the edge of the absorber in each case. For the plane absorber, proton fluence is increased just beyond the edge, with a corresponding decrease in fluence just beneath (Fig. 1, left). Further away from the edge, below the absorber, scatter equilibrium is established. In the case of the wedged absorber (Fig. 1, right), a single 'spike' in fluence is observed just beyond the 'toe' of the wedge. Protons scattered from the toe to the region just beyond the wedge are compensated for by greater scattering further within the wedged region.

Whilst aluminium wedges are used exclusively at our centre, other centres favour PMMA wedges. The choice of wedge material is influenced by scatter production, geometry and ease of

* Corresponding author. Radiotherapy Physics, The Royal Berkshire NHS Foundation Trust, Reading RG1 5AN, UK. Tel.: +44(0)118 322 8155; fax: +44(0)118 322 7377.
E-mail address: colin.baker@royalberkshire.nhs.uk (C. Baker).

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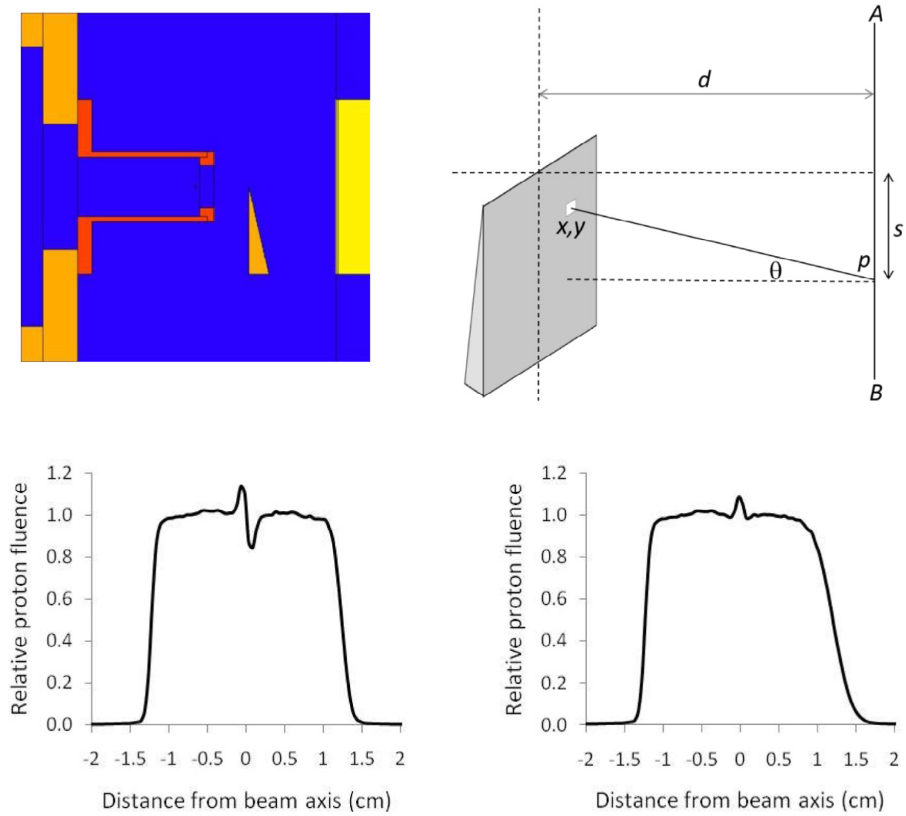

Figure 1. Top: Geometry of the treatment nozzle and wedge in Monte Carlo simulations (left) and schematic representation of the contributions to proton fluence at point, *p*, on profile A–B due to scatter in the wedge (right). Bottom: Proton fluence profiles along A–B for a 25 mm diameter, 60 MeV proton beam at 45 mm from a 1 mm Al slab (left) and 22.8° Al (40° in eye tissue) wedge (right), as predicted by Monte Carlo simulation.

manufacture. The aim of the wedge to is variably range-shift the proton beam with minimal scatter production. The ratio of scattering power to stopping power (rad^2 MeV^{-1}) increases approximately linearly with atomic number (Z) (compare for instance expression 24 in Gottschalk [5] for proton mass scattering power with expression 2.1 in Reference [6] for proton mass stopping power), so ideally a low-Z material, such as PMMA, would be preferred for range-shifting. However, the associated reduction in density in using PMMA over aluminium results in an increased thickness of the physical wedge, which can then be a limiting factor in wedge placement (the proximity of the nozzle to the patient). The proportional increase in physical thickness in moving from Al to PMMA can be estimated via ratios of density and stopping power; $(\rho_{Al}/\rho_{PMMA})(S_{Al}/S_{PMMA}) \sim 1.79$ for 60 MeV protons. In addition, the density of PMMA is subject to greater variation than that of metals and accurate machining is more difficult.

Methods

In this work, Monte Carlo (MC) simulations and radiochromic film (EBT2) measurements were performed on the CCC proton beamline, which has been treating ocular tumours since 1989. The proton beam is produced by a Scanditronix MC-62 isochronous cyclotron, initially intended for fast neutron therapy trials. The basic beamline construction has been described earlier [7] and more recently in References [8,9] where the use of wedges is illustrated by EYEPLAN [10,11] treatment planning software.

The geometry of the therapy nozzle, final collimator, wedge and patient position is shown in Fig. 1. Wedge angles (in Al) of 13.1, 22.8 and 40.9 degrees (corresponding to eye tissue wedge angles of nominally 25, 40, and 60 degrees) were placed with their edges close to the central axis of the beam and the resulting dose in a plane parallel to the wedged direction investigated for varying geometries both by Monte Carlo (MC) simulation and direct measurement using radiochromic (GafChromic EBT2) film. The distance of the tumour centre (isocentre) to the collimator is 70 mm at the CCC. Additional simulations and film measurements were performed using a 37° PMMA wedge (designed to give the same eye-tissue angle as the 22.8° aluminium wedge) for comparison.

Wedges are typically used only when treating relatively deep ocular tumours, in order to spare normal tissue such as the optic nerve or macula at the back of the eye. Simulations have therefore been restricted to full-energy beams (62 MeV incident on the beamline and approximately 60 MeV exiting the beam nozzle).

Monte Carlo (MC) simulations

Modelling of the Clatterbridge proton facility has been described previously [12,13]. MCNPX v2.7e was used to model the full beam-line including scattering foils and beam stopper, modulator wheel, nozzle, collimator and wedge. In this work the proton spectrum incident on the beamline was represented by a double-Gaussian function, fitted to reproduce the full-energy Bragg peak in water measured by a Markus ionisation chamber. The double-Gaussian source was found to provide an improved fit to measured depth-dose data in comparison to a single Gaussian used previously. In order to generate a clinical spread-out-Bragg peak (SOBP) in a single MC run, proton histories stored in a phase-space file (PSF) generated at the central position of the modulator wheel, in its absence, are subsequently re-started at appropriate depths in a solid block of PMMA according to the weightings of the true modulator vane angles (of up to 32 increments). The validity of this approach