PAMELA OVERVIEW: DESIGN GOALS AND PRINCIPLES*

Ken Peach#, John Cobb, Suzanne Sheehy, Holger Witte, Takeichiro Yokoi (JAI, Oxford), Richard Fenning, Akram Khan (Brunel University, Middlesex, UK), Rebecca Seviour (Cockcroft Institute, Lancaster, UK), Carol Johnstone (Fermilab, Batavia, USA), Mark Hill, Bleddyn Jones, Borivoj Vojnovic (Gray Institute for Radiation Oncology and Biology, Oxford, UK), Morteza Aslaninejad, Matt Easton, Jaroslaw Pasternak (Imperial College, London, UK), Jürgen Pozimski (Imperial College London and STFC/RAL, Chilton, Oxon, UK), Neil Bliss, Carl Beard, Peter McIntosh, Susan Smith, Stephan Tzenov (STFC/DL, Daresbury, Warrington, UK), Robert Edgecock, David Kelliher, Shinji Machida, James Rochford (STFC/RAL, Chilton, Oxon, UK), Roger Barlow, Hywel Owen, Sam Tygier (University of Manchester, Manchester, UK)

Abstract

The PAMELA (Particle Accelerator for MEdical Applications) project is to design an accelerator for proton and light ion therapy using non-scaling Fixed Field Alternating Gradient (ns-FFAG) accelerators, as part of the CONFORM project, which is also constructing the EMMA electron model of a non-scaling FFAG at Daresbury. This paper presents an overview of the PAMELA design, and a discussion of the design goals and the principles used to arrive at a preliminary specification of the accelerator.

INTRODUCTION

Charged Particle Therapy (CPT) [1] uses protons and light ions (for example, carbon) to treat some cancers. After being proposed [2] by R.R. Wilson in 1946, the first patients were treated with protons in Berkeley in 1954. CPT has now moved out of the laboratory and into the hospital, and there are now 29 facilities world-wide in operation with several more under construction. Over 70,000 patients [3] have been treated so far. Existing facilities use cyclotrons or synchrotrons to accelerate the charged particles. However, it is possible that improved performance could be achieved through the use of non-scaling FFAG accelerators, leading to faster energy variation and spot-scanning capability, contributing to a better patient experience, shorter treatment times and lower overall cost.

CLINICAL REQUREMENTS

The advantages of charged particle therapy over conventional radiotherapy using MV X-rays can be seen from Figure 1. At a single proton energy setting, a high dose can be delivered to the tumour while delivering a low dose to the healthy tissue on entry, and no dose on exit, whereas a single X-ray beam delivers a higher dose to the healthy tissue on entry, and a significant dose on exit. Of course, in conventional X-ray therapy, a lethal dose is delivered to the tumour while delivering a much lower dose to healthy tissue by irradiating the tumour from many directions. However, this has the effect of exposing a large volume of healthy tissue to radiation, including some potentially radio-sensitive organs. Using the same technique with protons results in a very significant reduction (between one half and one tenth typically) in the volume of healthy tissue irradiated, and/or a significant reduction in the dose received by healthy tissue. Light ions have a higher Relative Biological Effectiveness (RBE) – for carbon this is typically 3-5 – leading to an even larger ratio between the peak and the entry dose, but with a small fragmentation dose behind the tumour.

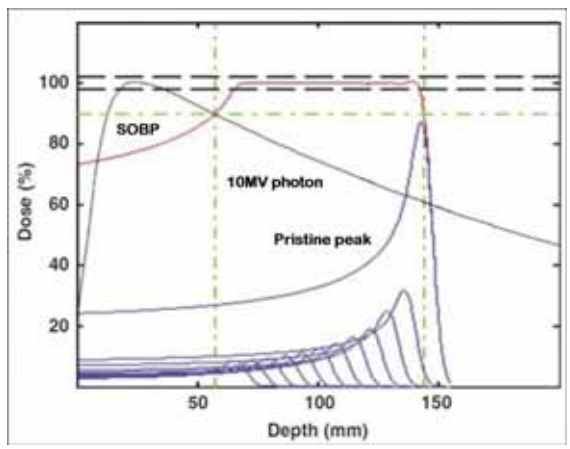

Figure 1: The % energy deposition as function of depth for protons and photons, showing the Bragg peak (the "pristine" peak), and the Spread-out Bragg Peak (SOBP). (from [1]).

For large tumours, the full tumour volume can be irradiated uniformly by varying the energy, so that the accumulation of the Bragg peaks takes account of the energy deposited in the tumour before the Bragg peak for the higher energy positions. As can be seen from Figure 1, the dose per volume element (voxel) has a large dynamic range (for a 10 cm tumour, this is about 20). The dose delivered to the tumour must be within 2% of the planned value.

A partial set of requirements on the accelerator capabilities for a clinical therapy system is shown in Table 1. The potential advantages of FFAG accelerators over cyclotrons and synchrotrons are that in principle they should be able to extract the beam at variable energy and at a high (~kHz) repetition rate, with the ability to change between protons and light

* Work supported by the UK Basic Technology Fund grant number EPSRC EP/E032869/1 # Ken.Peach@adams-institute.ac.uk

ions relatively quickly, and to be able to match cyclotrons and synchrotrons in terms of dose rate, dose stability and dose precision.

Table 1: A summary of the main clinical requirements on the accelerator for a Charged Particle Therapy centre with both protons and carbon ions.

Parameter		Value
Extraction energy (proton) [Min, Max]	[MeV]	60, 250
Extraction energy (carbon) [Min, Max]	[MeV/u]	110, 450
Energy step (proton) [@Min, @Max]	[MeV]	5, 1
Energy step (carbon) [@Min, @Max]	[MeV/u]	15, 6
Energy resolution (FWHM) [@Min, @Max]	[%]	3.5, 1.8
Voxel Size [Min, Max]	[mm]	4×4×4 10×10×10
Smallest Field of view [Min, Max]	[mm]	100×100 250×250
Dose rate (proton) [Min, Max]	[Gy/min]	2, >10
Dose rate (carbon) [Min, Max]	[Gy/min]	2, >10
Cycle rate [Min, Max]	[kHz]	~1
Bunch charge (proton)	[pC]	1.6
Bunch charge (carbon)	[pC]	0.3
Bunch charge stability	[%]	<10

PAMELA

The ns-FFAG was invented in 1999 [4]. In their original incarnation, the magnetic design was arranged to greatly compress the range of orbit radii and thus the magnet aperture, while maintaining a linear magnetic field, leading to expectations of smaller apertures, and thus significant cost reduction when compared with scaling machines. EMMA, the Electron Model with Many Applications will demonstrate the feasibility of this technology, and is described elsewhere [5]. Briefly, EMMA is a 42-cell, densely-packed ring, with the linear magnetic fields provided by displaced quadrupoles, and achieving rapid acceleration by using 19 1.3GHz cavities, each with an accelerating voltage of 20-120kV, giving an energy gain per turn of between 0.38 MeV and 1.28 MeV.

While this lattice is a natural starting point for PAMELA, there are features that make it unsuitable for protons and light ions. Studies with a 48-cell densely packed linear lattice showed that it was difficult to achieve the high packing fraction with a realistic magnet design, and there was insufficient space in the short straight sections for the variable frequency RF cavities needed for the non-relativistic acceleration. It was also shown that the requirements on the field accuracy and alignment precision were severe.

An alternative is to study a less dense lattice with longer straight sections, which means departing from simple linear magnetic fields. An advantage of this approach is that it is then possible to stabilise the horizontal and vertical tune (to avoid resonance crossing) and to limit the orbit excursion. There are several ways to achieve this. One approach [6] uses both edge and alternating gradient focussing to stabilise the tune. An alternative approach [7] is to stabilise the tunes through the addition of higher-order multipoles. One such lattice is shown in Figure 2.

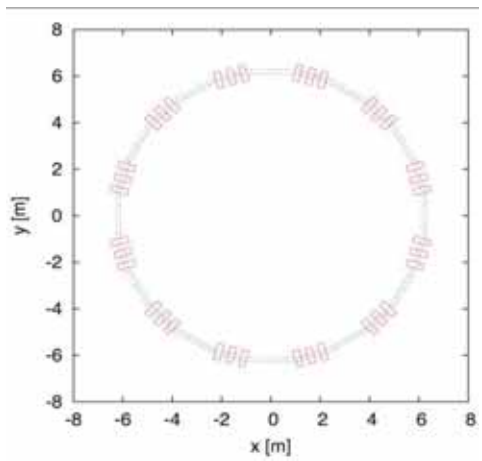

Figure 2: A tune-stabilized non-linear non-scaling FFAG lattice for protons 31 MeV to 250 MeV.

The lattice has 12 triplet (FDF) cells, with a median radius of 6.25m and 1.95m long straight sections (about 1.7m of useful length) see Table 2.

Table 2: Lattice Parameters

	Injec-tion	Refer-ence	Extrac-tion
Proton K.E. [MeV]	31	118	250
C^{6+} K.E [MeV/u]	7.8	31	68.4
Bρ [Tm]	0.81	1.62	2.43
# Cells, R_0 [m]	12, 6.25		
K value, D/F ratio	38, 1.35		
B^D_0, B^F_0 [T]	2.25, 1.67		
Packing factor	0.48		
Long, Short drift [m]	1.7, 0.31		
Magnet length [m]	0.31		
Orbit excursion [m]	0.17		

The performance of this lattice is discussed in [8]. Preliminary ideas for the design of the magnets [9] and the RF [10] have been developed. The field shapes for the main ring magnets [9] are shown in Figure 3. An outline of the RF cavity from reference [10] is shown in Figure 4, with the principal parameters in Table 3.

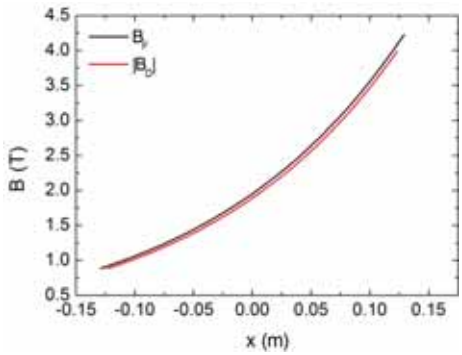

Figure 3: Magnetic field as a function of radius for the F and D magnets.

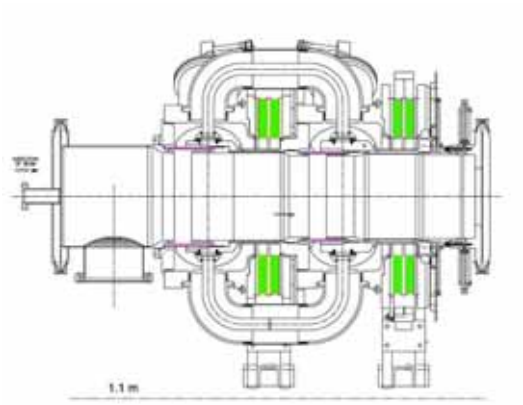

Figure 4: PAMELA RF cavity

Table 3: RF Cavity Parameters

Frequency [h=10] injection	[MHz]	19.4
Frequency [h=10] extraction (max)	[MHz]	46.2
Repetition Rate	[kHz]	1
Energy gain/turn	[keV]	100
Number of cavities		≤ 8
Length	[mm]	1100
Aperture	[mm]	230

There are also preliminary ideas ([11]-[12]) for the injection chain (see Figure 5). The protons and carbon ions will be produced in separate sources, allowing faster switching between ion species in a clinical situation, improving the productivity of the facility. A Low Energy Beam Transport line (LEBT) will transport the particles from the sources into a pre-accelerator, and another beam transport section (MEBT) will inject the particles into PAMELA. A standard 30MeV proton cyclotron can be used for the proton beam injection, and a radio frequency quadrupole (RFQ) and linac can be designed for the carbon injection.

Finally, in order to achieve the performance requirements in the treatment room, it is necessary to use an achromatic beam transport and gantry. Studies are under way [12] to design an FFAG-like beam transport system.

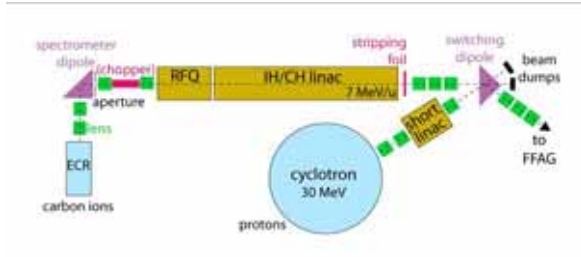

Figure 5: Schematic of proposed injector assembly, including ion sources, LEBT, pre-accelerators and MEBT. The proton source is contained within the cyclotron.

REFERENCES

[1] See, for example, W.P. Levin *et al*, "Proton beam therapy", British journal of Cancer 93 (2005) 849-854 for a review of Charged Particle Therapy.

[2] R.R. Wilson, "Radiological uses of fast protons", Radiology 47 (1946) 487.

[3] M. Jermann, "Hadron Therapy Patient Statistics", March 2009, http://ptcog.web.psi.ch/.

[4] C. Johnstone et al, "Fixed Field Circular Accelerator Designs", PAC'99, New York, March 1999, p. 3068.

[5] S.L. Smith, "EMMA, the World's First Non-Scaling FFAG Accelerator", WE4PBI01, this conference.

[6] C. Johnstone and S. Koscielniak, "Tune-Stabilized Linear-Field FFAG for Carbon Therapy", Proceedings of EPAC Edinburgh (2006), 2290.

[7] S. Machida, "FFAGs for proton acceleration", FFAG'08 workshop, Manchester, 2008, http://www.cockcroft.ac.uk/events/FFAG08/presentations/Machida/machida_proton_0902.pdf;
S. Machida, "Scaling fixed field alternating gradient accelerators with small orbit excursion", submitted to Phys. Rev. Lett.

[8] S. Sheehy et al, "PAMELA Lattice Design and Performance", FR5PFP001, this conference.

[9] H. Witte et al, "PAMELA Magnets - Design and Performance", MO6PFP073, this conference.

[10] T. Yokoi et al, "PAMELA: Development of the RF System for a Non-Relativistic Non-Scaling FFAG", WE5PFP011, this conference.

[11] M. Aslaninejad et al, "Injection Layout for PAMELA", MO6RFP029, this conference;
M. Easton et al, "RFQ Design Optimisation for PAMELA Injector", FR5REP066, this conference.

[12] R. Fenning et al, "An FFAG Transport Line for the PAMELA Project", TH6PFP022, this conference.