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REVIEW

Back to the future: the history and development of the clinical linear accelerator

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Abstract

The linear accelerator (linac) is the accepted workhorse in radiotherapy in 2006. The first medical linac treated its first patient, in London, in 1953, so the use of these machines in clinical practice has been almost co-existent with the lifetime of *Physics in Medicine and Biology*. This review is a personal selection of things the authors feel are interesting in the history, particularly the early history, and development of clinical linacs. A brief look into the future is also given. One significant theme throughout is the continuity of ideas, building on previous experience. We hope the review might re-connect younger radiotherapy physicists in particular with some of the history and emphasize the continual need, in any human activity, to remain aware of the past, in order to make best use of past experience when taking decisions in the present.

History is a picture gallery in which there are a few originals and many copies: Alexis de Tocqueville, L' Ancien Régime (1856).

(Some figures in this article are in colour only in the electronic version)

1. Introduction: aims, scope and limitations of this review

Installation of the first clinical linear accelerator began in June 1952 in the Medical Research Council (MRC) Radiotherapeutic Research Unit at the Hammersmith Hospital, London. It was handed over for physics and other testing in February 1953 and began to treat patients on 7 September that year. Thus the linac's progress to becoming the dominant machine in modern radiotherapy has been almost simultaneous with the development of *Physics in Medicine and Biology* (PMB). In line with the remit given for contributions to this anniversary issue of PMB, this review is intended as the authors' personal perspective on developments in the field over the journal's lifetime, leavened with relevant personal statements. Unlike many of the other reviews in this issue, and also unlike their distinguished authors, we were not involved in the early development of our subject, as the timescale is long enough not to have allowed that! However we have had contact with many of the early contributors and we have both worked a lot with clinical linacs! One of us (JBT) entered RT physics in the Glasgow department in the 1960s, moved to Leeds in the early 1970s and retired just as this review was requested. During this career, he worked with machines from AEI (Metropolitan-Vickers), Mullard Equipment Ltd (MEL), Radiation Dynamics Limited (RDL), Philips, Elekta and Varian as well as with a range of Co units. The other (DIT), having worked with a very temperamental van de Graaff ion accelerator during his PhD, joined the Edinburgh radiotherapy centre in the early 1980s and whilst there has worked on linacs from RDL, Brown Boveri Corporation (BBC, later Asea Brown Boveri, ABB), and then Varian and since 2005 mainly with Elekta linacs in the Yorkshire Cancer Centre in Leeds. Both authors have been involved in British Standards Institute (BSI) committees on radiotherapy equipment and have contributed to International Electrotechnical Commission (IEC) recommendations on linac safety and performance standards.

As there are a range of textbooks and review papers already available which include thorough descriptions and discussions of linac design, theory, history, developments and current technology, we feel no need to repeat a lot of that detail here. Therefore we have included a relatively small number of references for a review paper. However, we do refer the reader requiring more detail to some of those other wide-ranging sources. In addition, the review was requested just as one of us retired and the other had not long before moved from one centre to another, so it seemed an appropriate time to look back and reflect, to take the editor's injunction to heart and to indulge ourselves in a nostalgic account of the subject. Thus the review is a personal selection of things we feel are interesting in the history, particularly the early history, and development of clinical linacs. We hope it might re-connect younger radiotherapy physicists in particular with some of that history. If we omit to mention some publications or give them less emphasis than others might, we ask that this is viewed in this context.

2. Review of PMB's role and some other sources

Given the nature of this anniversary issue, we have considered PMB's role in reporting relevant developments. Therefore we recognize that we might be giving a rather UK-centred view. However the early development of linacs and their first clinical implementation in the UK were also closely paralleled in the US, with many other countries soon following suit as regards clinical implementation and a few others as regards linac manufacture. The widespread clinical use of linacs has ensured that developmental ideas have come from many sources and it may be noted that the papers in PMB concerning linacs clearly reflect that, bearing out the journal's robustly international nature.

It is impossible to divorce developments and publications relating to linacs from other developments going on in radiotherapy at the same time. These include, among others, closely inter-related advances in: imaging; dosimetry; treatment planning methods, systems and algorithms; plan evaluation and optimization; beam and high dose volume shaping; information systems and flows; connectivity; immobilization and positioning; verification; process safety; etc. Nor can they be separated from clinical developments, particularly of treatment techniques, as these are spurred by increased cabability and functionality of linacs, whilst in turn linac improvements are driven by the demands of clinical and associated system developments (as well as commercial imperative and competition, of course!). Therefore the selection of which papers are relevant is subjective and may reflect how widely or not the boundaries are drawn around the topic. Thus there is clear overlap with other reviews in the area of radiotherapy in this issue of PMB, particularly with those on IMRT (Bortfeld 2006, Webb 2006), electron beam therapy physics (Hogstrom and Almond 2006) and tomotherapy (Mackie 2006).

Bearing this in mind whilst looking at PMB's contents over 50 years provides some interesting observations. There were very few specifically relevant papers before 1964, although there was a wide-ranging state-of-the-art review of supervoltage therapy by Frank Farmer from Newcastle in April 1962. (In passing it may be noted that the first advert in PMB for the Baldwin-Farmer dosemeter, forerunner of modern 'Farmer-design' based ionization chambers and dosemeters, appeared in November 1957.) In contrast, the *British Journal of Radiology* (BJR) carried a significant number of papers describing linac design and clinical implementation over those years. Seminal works also appeared in *Nature* (e.g. Miller (1953)) and in the *Proceedings of the Institution of Electrical Engineers* (e.g. Miller (1954)). Other than Farmer's 1962 review, PMB can only be used to track the development of high energy machines in that period by the adverts it contained. By 1962 those for linacs were beginning to oust those for van de Graaff and other machines (although of course adverts for Canadian and UK Co units still featured strongly, including the 'Hunslet' machines manufactured in Leeds). An advert for Mullard linacs from November 1962 states that they are 'now in use in three continents'. Mackie's (2006) review provides an evocative echo of this by using the same phrase to describe the current status of tomotherapy units.

In the 1960s PMB carried around 20 directly relevant papers; topics or themes can be identified as including dosimetry, beam spectra, tissue compensators and other accessories. In the 1970s and 1980s approximately 25 relevant papers appeared per decade. Topics in the 1970s include electron beams, photo-neutrons, coordinate systems, accelerator dosemeter design, lasers for positioning and back-pointing and dosimetry. Topics and themes in the 80s include dosimetry, photo-neutrons, conformal radiotherapy, geometric and dosimetric accuracy, computer-assisted set-up, independent jaws and the first mention of dynamic MLC. In the early 1990s an average of around 5 relevant papers per year appeared, rising to 15 or so per year on average in the later 1990s and 20 or so per year since 2000. A multiplicity of themes can be listed, reflecting rapid progress in many areas. In the 1990s, these include asymmetric collimators, MLC, dynamic wedge, dynamic MLC, IMRT, IMAT, stereotactic systems, EPIDs, megavoltage CT, cone-beam kV CT, robotic and tomotherapy approaches, Monte Carlo modelling of linacs, spectra, safety, the ubiquitous dosimetry again, and more. Since 2000, most of these themes have continued and others have been added, including, for example, IGRT, respiratory-gated systems and approaches, patient support systems design, etc. As examples of two very recent papers, Vassiliev *et al* (2006) discuss beams produced with the flattening filter removed for IMRT use, following on from Fu *et al* (2004), and Loi *et al* (2006) discuss neutron production from a mobile linac used for intraoperative electron therapy.

Many papers could be cited as meriting attention. However we list only a few here, selected as noteworthy from the advantage of our current perspective and interests. Benner *et al* (1962) produced an early paper on fluorescent screen and TV-camera based portal imaging. The first dosimetry protocol for MV x-rays appeared in 1964, produced by the Hospital Physicists Association (HPA 1964), forerunner of the present-day Institute of Physics and Engineering in Medicine (IPEM). One of the first international dosimetry intercomparisons involving at least some linac-produced MV x-rays (Almond *et al* 1972), was reported by three workers that one of us (DIT) has worked closely with on many occasions and which was a forerunner to many other national and international dosimetric intercomparisons (e.g. Thwaites *et al* 1992, Nisbet and Thwaites 1997). Brahme *et al* (1982) published the paper now recognized as defining the inverse planning problem in IMRT. Kallmann *et al* (1988) provided probably the earliest paper on delivery of arbitrary dose distributions by dynamic MLC. In 1985, Amsterdam's development of the matrix liquid ion chamber EPID appeared (Meertens *et al* 1990), whilst 1986 saw a paper on the possibilities for transit dosimetry and on-line verification using imaging (Leong 1986). Convery and Rosenbloom (1992) published the basis of the sweeping leaf technique for dynamic IMRT delivery and for its optimization. In 1990 the first dosimetry code of practice based on direct provision of absorbed-dose-to-water calibration factors across a range of MV photon energies was published (HPA 1990); followed in 2003 by a similar protocol for electrons (IPEM 2003). These were linked to utilization of the UK National Physical Laboratory's (NPL) pioneering calorimeter-based calibration services. However the proposal that therapy beam calibrations should be based on dosimeter calibrations in terms of dose to water went back at least to 1979 (Reich 1979). In 1995, Cho *et al* discussed cone beam kV CT for radiotherapy, at that time for stimulators, but giving a technique that was later utilised in the William Beaumont Hospital development of x-ray volumetric imaging for linac-mounted systems (Jaffray *et al* 1999, Letourneau *et al* 2005). Finally, as a tribute to close colleagues of one of us (DIT) and in appreciation of the time and effort saved in linac commissioning and QA by the introduction and widespread use of these systems, probably the first paper on a computer-controlled beam data acquisition system was given in Bottrill *et al* (1975).

PMB has previously published reviews covering linac technology and development (Farmer 1962, Karzmark and Pering 1973). Other reviews have also touched on the subject, e.g. Meredith (1984) in an issue reviewing 40 years of medical physics, to mark the 40th anniversary of the founding of the Hospital Physicists Association. In addition, elsewhere than PMB, noteworthy reviews have been given by Karzmark and collaborators (e.g. 1984) and Ginzton and Nunan (1984). Also many books are useful, either for directly covering the topic or for covering related practical areas, such as specification, dosimetry, commissioning and QA; the authors would specifically like to draw attention to Greene and Williams (1997), Karzmark *et al* (1993), Klevenhagen (1983), Metcalfe *et al* (1997), Van Dyk (1999, 2005), and Williams and Thwaites (2000) for further reference.

As a final cautionary note from the trawl through 50 years of PMB, the July 1962 issue contained a stern warning on plagiarism, so the authors have attempted to be very careful in preparing this review! However, reading so many superb historical sources in this process may have lead to the use of similar wording in homage to one or other of the giants on whose shoulders we stand!

3. Historical overview of the early clinical implementation of linacs

The advantages of higher energy radiotherapy beams had been recognized from early in the development of the speciality, but appropriate equipment was lacking. However megavoltage external beam radiotherapy began in 1937, just as radiotherapy was moving from being mainly a palliative procedure to being a major curative agent in cancer treatment and just as it was moving to a reproducibly systematic and quantitative basis (Meredith 1984). In that year, treatments began at St Bartholomew's Hospital, London, using a 1 MV x-ray unit having two symmetrically arranged Cockcroft-Walton 500 kV generators and a 30 ft (approx. 9.25 m) long x-ray tube to achieve the magic number (Allibone *et al* 1939). This machine, designed and built by the Metropolitan-Vickers Electrical Co Ltd, incorporated many advances, even though based on conventional acceleration techniques. In 1987, at a special BIR meeting at St Bartholomew's to commemorate the 50 year anniversary of that historic beginning, George Innes gave an engaging account of the physicist's testing and experience of this unit (later published in Innes (1988)). In his telling it involved frequent small doses of radiation, large electric shocks, especially if turning one's back on the machine on damp days, and enormous enthusiasm for the task. As George lived well into his 90s, this may be a small fragment of evidence for the hormesis effect, although we concede it is firmly anecdotal! Also in 1937, the roentgen was re-defined and was accepted as the unit of radiation quantity; beam direction devices had recently been developed; and wedge filters and beam flattening filters were soon to be available (Meredith 1984). Thus the stage was set for higher energy machines and awaited development of appropriate technology. Whilst a range of approaches were tried, the linear accelerator has proved to be the system of choice as viewed from today's perspective and its first faltering clinical steps were taken by the Hammersmith machine.

In 1953 this linac, developed and built by Metropolitan-Vickers (Met-Vic) and using a 3 m long accelerating or 'corrugated' waveguide to produce 8 MV x-rays, was the only one in the world treating patients. Shortly after that two 4 MeV machines were installed, one in Newcastle by Mullard Equipment Ltd, a division of Philips, and one in the Christie Hospital, Manchester by Met-Vic. The latter company supplied two more clinical 4 MeV machines (by then known as Orthotrons) in 1955 to the Western General Hospital, Edinburgh, and to Mount Vernon Hospital, Northwood, Middlesex (Miller 1956). Meanwhile the research group at Stanford University, California, USA had developed a 6 MeV clinical linac which was installed in the Stanford Department of Radiology in 1954, treating its first patient in January 1956, at which time there were seven clinical linacs in the world. A few other units had either been installed or were under installation at that time in centres in the UK, including a 15 MeV unit at St Bartholomew's Hospital, and in the US, in Chicago, but it is not clear that they were yet clinical (Miller 1956, Karzmark and Pering 1973). Three Met-Vic units became operational in Australia in 1956–1957 (Brisbane, Melbourne, Adelaide) with one soon after in New Zealand, whilst other units were supplied by Mullard to Australia, Japan and Russia by the time of their 1962 advert quoted above. Varian installed its first prototype 6 MV fully isocentric linac in UCLA Medical Center in 1962. The first Mevatron (then from Applied Radiation) was installed in 1965; the first Sagittaire from CSF in Paris in 1967 and the first Toshiba in 1969 (Karzmark and Pering 1973). At the time of Farmer's 1962 review, there were 15 clinical linacs worldwide (as well as 'many' van der Graaff machines and 50 betatrons); by 1968 79 linacs, as compared with 20 van der Graaff, 137 betatrons and around 1700 Co units (IAEA 1968). By Karzmark's (1984) review, he could state that over one-half of all US MV treatment units were linacs and over 90% of new units. As linacs have established themselves as the machine of choice, their numbers have grown well into the thousands (e.g. Podgorsak *et al* (1999) quote Varian's production numbers up to that date as 3200 and it is likely that total production from the other manufacturers was of a similar size). Co unit numbers worldwide have been larger overall, but Co units have been largely replaced by linacs over the last 25 years or so in developed countries and are now increasingly likely to be overtaken by linacs elsewhere, as new centres are established and older units are replaced.

4. A note on linacs' family trees

A number of commercial companies have been mentioned, which might be confusing to those who believe there might have only been three manufacturers! Apparently the clearest and longest single name thread for linacs is Varian, set up by the Varian brothers. The company began producing clinical linacs in 1962 with that name and remains with it in 2006. However, the current Elekta machines have a longer history in a single manufacturing base in Crawley, UK, albeit under three names! They were originally sold under the Mullard Equipment Ltd banner and then directly under the Philips name. Later the Philips radiotherapy product division became part of the Elekta Oncology Systems group. Meanwhile Siemens bought out other companies and designs for the initial basis of its linac production, but was of course a long-established name in medical radiation products, including betatron production. The earliest UK manufacturer, Metropolitan-Vickers, was already a successful company making kV units, when it developed and produced the first clinical linac in the early 1950s. It later

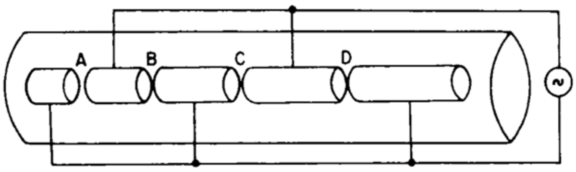

Figure 1. The 'drift-tube' linear accelerator.

became AEI, then the radiotherapy division became Radiation Dynamics Limited of Swindon, UK. The company, under-capitalized for developments in a very competitive market, sold its 'Dynaray' designs to the Brown-Boveri Corporation of Switzerland at the beginning of the 1990s, as part of the latter's strategy to switch from betatron to linac production. They brought with them a strong history in computer control of other products which they added to the solid RDL linac design and, amongst other innovations, worked with the Amsterdam group to commercialize the liquid ionization matrix EPID. The radiotherapy division then merged with a Swedish company to become Asea Brown Boveri. These machines sold well in Europe, but before they entered the N American market, the company was bought by Varian. It is interesting to note that cross-links continue, e.g. IMPAC, the oncology information systems supplier originally begun by Varian software engineers has recently merged with Elekta. There have been many other manufacturers in the field from a number of countries, including Canada, China, France, Japan, Poland, Russia, USA, etc and also Swedish manufacturers of microtrons and racetrack microtrons. However only a few have survived the bracing competition, whilst at the same time continuing to support and fund the required rapid and escalating technological product development in the field.

It must also be noted here that other companies have developed revolutionary new concepts, either as part systems or complete machines, noteworthy examples including the Nomos Corporation's add-on serial tomotherapy system, Tomotherapy Inc's helical tomotherapy machines (Mackie 2006), Accuray, Novalis and others' robotic arm machines, and a range of companies making specific stereotactic solutions, microMLCs, and other systems.

5. Origins of the linear accelerator

The history of the linear accelerator goes back further than clinical implementation. Following an earlier proposal by the Swedish physicist Ising, the concept was developed in a 1928 paper by Wideröe, a Norwegian physicist working for the Brown-Boveri Company of Switzerland, under the very modest title 'On a new principle for the production of higher voltages'. This was the forerunner of the 'drift-tube' type of linear accelerator (figure 1) consisting of a series of co-linear tubes connected alternately to opposite ends of a high-frequency HT generator. The particles experience an energy gain each time they cross a gap. If the ac frequency is constant, the tubes must progressively increase in length to ensure that the particles always arrive at each gap at the correct point in the cycle as their velocity increases. Therefore, although the drift-tube linac came to be widely used in experimental physics both as an accelerator in its own right and as an injector for higher energy machines, it was suited only to the acceleration of heavier particles where the increase in velocity is relatively modest. For electrons, because of their small mass, the velocity increases so rapidly as to make the system impractical, so it was not useful for medical applications requiring the production of high energy x-rays from electron acceleration. It may be noted also that in the same paper, Wideröe (1928) proposed the 'beam transformer', the basis of the magnetic induction part of the betatron.

6. The development of the microwave linear accelerator

It would not have been obvious to the workers developing microwave technology, much of it linked to military radar requirements between 1935–1945, that their work would provide the basis for equipment that would revolutionize radiotherapy. In the mid-1930s Hansen in Stanford (Hansen 1938) developed the concept of an electron accelerator based on the principle of passing electrons repeatedly through a resonant microwave cavity, gaining an increment of energy on each pass, the system being termed the 'Rhumbatron'. However the power levels available from the microwave generators of the time were woefully inadequate for practical purposes. As with many other branches of technology the needs of World War II were to provide the impetus for the necessary developments. Wartime radar required microwave generators with MW outputs and two tubes were developed with such capabilities in 1939; the magnetron devised by Boot and Randall (1976) in the UK and the klystron by the Varian brothers (Varian and Varian 1939) in the USA. The fundamental difference is that the magnetron is a self-oscillator, producing oscillations in response to a dc input, while the klystron is essentially an amplifier with a low-power microwave input. The wartime radar systems were operated at a frequency of 3 GHz corresponding to a free-space wavelength of 10 cm and fortunately this was well suited to its subsequent adoption in electron accelerators.

The development of the microwave linac after 1945 took place independently on both sides of the Atlantic. One group under D W Fry at the Atomic Energy Research Establishment (AERE), then based at the Telecommunications Research Establishment at Great Malvern, England, and one under W W Hansen at Stanford University in California. By all accounts the two groups had little knowledge of the other's work until the late 1940s when after leapfrogging each other's achievements Fry's group had produced a clinically workable 8 MeV design (Miller 1954, 1955) and Hansen's an energy of 6 MeV (Ginzton *et al* 1948, 1957). The underlying excitement is almost palpable in some of the early contemporaneous accounts, even though still written in measured scientific style. Both systems employed the travelling-wave principle. Whilst this is briefly described below, a full treatment of the theory of microwave linacs can be found in Greene and Williams (1997), Karzmark and Pering (1973), Karzmark (1984), Karzmark *et al* (1993) and Podgorsak *et al* (1999).

7. Direct accelerators, other approaches and their comparison to linacs

By 1953 and over the next few years when the first linacs were beginning to treat patients, a range of other machines were available for the production of high energy x-rays. A number of direct accelerators were tried, i.e. those in which the energy of the emergent particles is directly related to (and limited by) the applied potential. Systems based on the Cockroft-Walton voltage multiplier and the van de Graaff electrostatic generator were produced, both proving to be relatively unsatisfactory either in terms of reliability and stability or lack of manoeuvrability. The principle of the resonant transformer was also employed; while acceptable for the production of orthovoltage quality x-rays, the bulky insulation required for megavoltage levels rendered the equipment somewhat unsuitable for clinical use. The illustration of a 4 MeV resonant transformer machine in figure 2 demonstrates this point. As summarized and compared by Farmer (1962), the significant alternatives to linacs were

- resonant transformers normally of up to 2 MV, requiring treatment rooms occupying more than two storeys in height;
- van de Graaff units also operating at 2 MV, whose major problems were mechanical, linked to the moving components for the high speed belt movement within a pressurized tank;

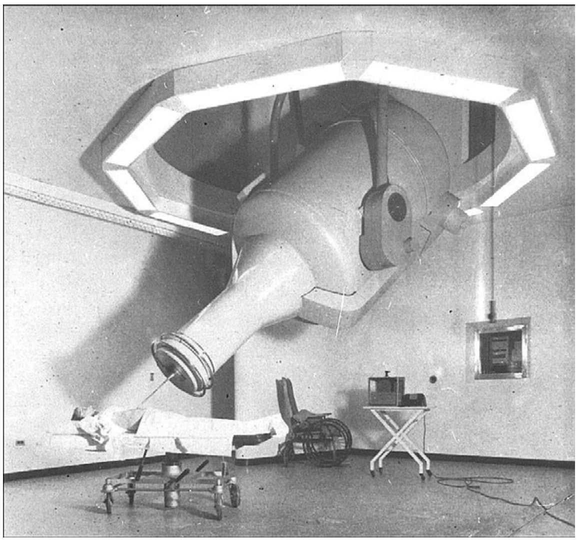

Figure 2. A 4 MeV resonant transformer unit.

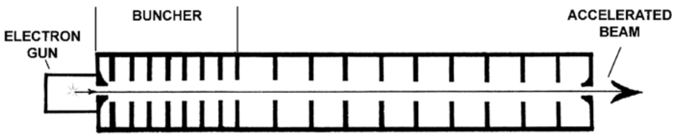

Figure 3. The travelling-wave accelerating structure.

- betatrons from Siemens, Allis-Chalmers and Brown-Boveri, of up to 35 MV (the latter manufacturer supplying a diagnostic x-ray tube incorporated in the machine for set-up, with the ability to transpose the therapy and diagnostic sources); the main problems being dose rate and stability;
- and, of course, the Co unit, the first of which had begun treatments in 1951.

It is interesting to consider some of Farmer's 1962 conclusions. He compared performance and costs of the systems. A 4 MV linac at £45 000 was roughly comparable to the cost of a betatron, but cheaper than a resonant transformer unit. However it was double the price of a Co unit or a van de Graaff machine. Nevertheless, when he considered throughput, costs per patient were similar for linacs and Co units and cheaper than for other high energy units. He concluded that linacs were the machine of choice for larger departments, with Co units possibly being the economic alternative for small centres. His view of the future for linacs was for simpler manufacture and more robust operation, better vacuum systems, reduced size of the machines and better mountings, all of which came to pass in a relatively short time. In fact the development of all of these features was significant in giving linacs the edge over other rival systems as the machine of choice for modern radiotherapy.

8. Basic concept of the travelling-wave linear accelerator

Figure 3 illustrates the basic travelling-wave system. Electrons are injected into a length of waveguide where they are accelerated by the axial electric component of the microwave

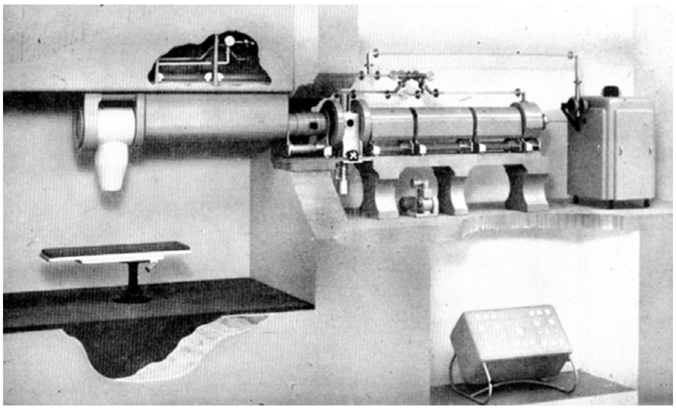

Figure 4. A model of the 1953 8 MeV linac installation at Hammersmith Hospital.

pattern travelling left to right. The plain cylindrical waveguide is modified by the insertion of irises, which have the effect of initially reducing the phase velocity of the waves to match the electrons' injection velocity, in order to 'capture' them. The phase velocity is then increased rapidly over the first section of the waveguide, as the electron velocity increases. This section is referred to as the 'buncher' and here the energy gain of the electrons is mainly kinetic. Then over the main length of the system the phase velocity is almost constant (just below c) while the electrons continue to gain energy but now mainly by a relativistic increase in mass. The 'buncher' is so called because, in traversing this section, the captured electrons progressively form into bunches near the crest of each advancing wave, where the efficiency of this bunching process is a figure of merit in waveguide design. (One of the first published uses of the often repeated analogy of electron bunches on microwaves being similar to a surfer was given by Miller (1953), although we feel it ought to be Californian in origin!) For a given design the final energy of the electrons is $\propto (LW)^{\frac{1}{2}}$ where W is the microwave power input and L is the length of the system. However the parameter which indicates the efficiency of a particular waveguide design, and which ultimately determines the length needed to produce a given energy with a particular level of power input, is the 'shunt impedance'. This is expressed in MΩ m^{-1} and is a measure of the energy gain per unit length per unit power input. While modern designs of accelerating waveguide can produce values in excess of 100 MΩ m^{-1} the first designs used for clinical machines struggled to achieve 20 MΩ m^{-1} and so a long waveguide was a feature of all the early machines.

9. Design of the first clinical machines

The first clinical linac at Hammersmith hospital, developed as a co-operative effort between AERE, MRC and Met-Vic, is illustrated in figure 4. The accelerating waveguide was long, at 3 m to produce 8 MeV, and was fed by a 2 MW magnetron. It was horizontal and the electron beam was deflected through 90° onto the target. Treatment fields of up to 20 cm square were possible at 1 m, with dose rates around 150 r min^{-1}. Although the linac itself was stationary, the treatment head could be swivelled through a limited angle and this feature, together with movements of the treatment couch, permitted some flexibility of the angle of the treatment beam directed at the patient. However, while it was still being installed, a new design was being developed to meet UK Ministry of Health requirements, to a basic specification by Howard-Flanders and Newbery (1950) at the MRC, and a number of these 4 MV linacs were

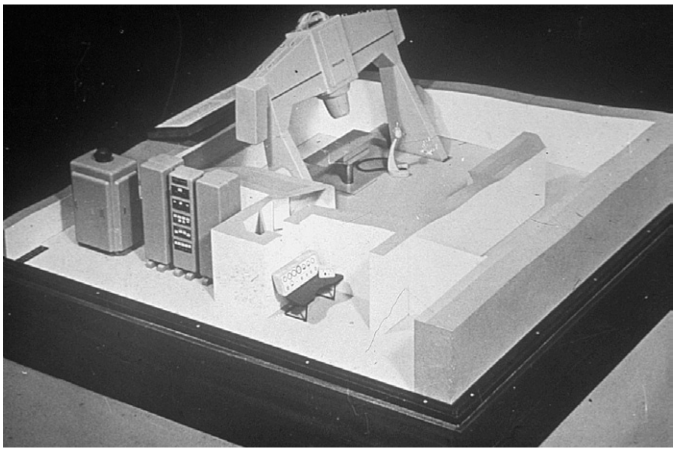

Figure 5. A model of the 4 MeV Mullard linac at Newcastle General Hospital.

Figure 6. A 4 MeV AEI Orthotron at Velindre Hospital, Cardiff.

ordered for other centres throughout the country. These were isocentrically mounted, however with limited angular movement. Two designs were developed to meet this specification; a double gantry mounted system devised by Mullard Laboratories (figure 5) and a single gantry from Metropolitan-Vickers (figure 6). The latter, for example, had a 1 m straight-through guide, operated at 200 r min^{-1} and could rotate to $\pm 120°$. The field size achievable was up to 25 cm $\times$ 30 cm (but with cut-off corners). Many features that we now take for granted were

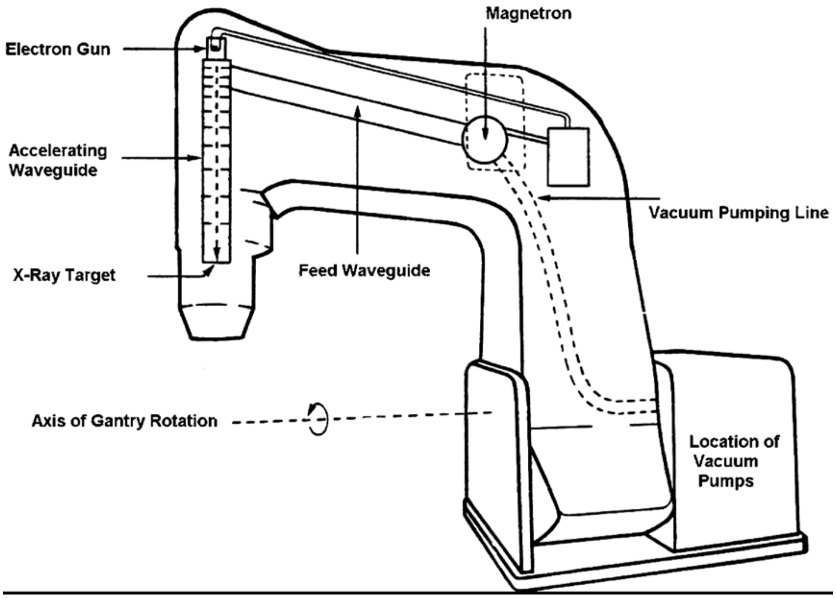

Figure 7. Configuration of major components in the 4 MeV Orthotron.

incorporated in this design, such as light field, wedges and diaphragms moving in an arc to produce improved penumbra.

In contrast to the UK developments under the auspices of official bodies, the design of the first clinical machine in the US (Ginzton *et al* 1957) was the result of a concept developed by H Kaplan, Head of Radiology at the Stanford University Medical Centre, and E Ginzton, a member of Hansen's research group at Stanford University. It was constructed in the Stanford University Microwave Laboratory, before being installed in Kaplan's department. The 6 MeV machine incorporated a 1.65 m long accelerating waveguide and was positioned in a trunnion mount originally designed for van de Graaff accelerators. As with the early UK designs this gave limited movement, but far short of the full isocentric rotation that was the ultimate objective of all the designers. Although it was the original intention that the power source would be a magnetron, during the time of the linac's development a 1 MW klystron had also been developed at Stanford (by the Varians) and was used to power the machine. One interesting feature was the incorporation of a 100 kVp x-ray tube which could be inserted near the target position to permit viewing the patient portal with an image intensifier, as also used with some of the betatrons in clinical use at around the same time. On board kV systems are not as new as we like to think!

10. Design limitations of the early machines

10.1. General

All machines in this first generation of clinical linacs exhibited relatively low beam energy and radiation output due mainly to inefficient waveguide design and limited microwave power from the magnetrons available at the time. However the most obvious limitation was the restricted range of movement. This was far short of the full isocentric rotation required for the ideal clinical machine. The sectional view of the Met-Vic Orthotron in figure 7 illustrates two reasons for this limitation.

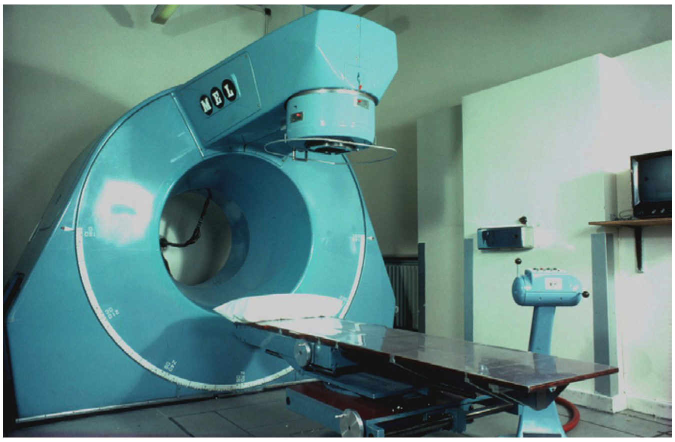

Figure 8. A 6 MeV MEL linac at Cookridge Hospital, Leeds, mid-1960s.

10.1.1. Vacuum pumping. The entire accelerating structure has to be held under high vacuum but the only appropriate type of pump available at the time was the oil-diffusion pump. As this has to remain stationary and vertical, the only option for designers was to locate it in the base support structure (together with its mechanical backing pump) and then connect to the accelerating structure through a long pumping line incorporating a rotating vacuum seal. This line had to be large diameter as its cross-section determines the pumping speed. The end result was that the total volume being continually pumped was far greater than the accelerating structure itself. An additional problem was that such diffusion pumps exhibit 'back-streaming' i.e. the migration of oil vapour back into the pumped volume which progressively contaminates surfaces right up into the accelerating structure itself. One of the authors (JBT) has vivid (and unwelcome!) memories of sitting late at night on the floor of a linac room surrounded by the dismantled components of the waveguide system, all nicely coated with a black film which had to be meticulously cleaned off by hand; all because of a fault in the diffusion pump system. This was radically changed by the development of the sputter-ion vacuum pump in the early 1960s, which had no working fluid and hence no possibility of contamination. It was compact and could be mounted in any orientation within the moving structure, close to the accelerating waveguide itself. It is no exaggeration to claim that the arrival of this type of vacuum pump liberated the designers of the clinical linac and allowed rapid evolution of practical flexible treatment units.

In contrast, in the first machines in the US, the use of oil-diffusion pumps was avoided by permanently sealing the entire accelerating structure under high vacuum but this practice had its own disadvantages and once the sputter-ion pump became available (the Vac Ion pump was a Varian invention) this was used.

10.1.2. Beam bending. The other dominant feature of the design in figure 7 was the size of the structure, the top of the gantry being some 4 m from the floor. This was due to the size and position of the accelerating waveguide. However, mounting the waveguide horizontally, with the beam being deflected magnetically before striking the target and with the vacuum pump positioned within the moving structure, enabled a radically different configuration. Figure 8 shows one of the first of the next generation of clinical linacs embodying these changes (an MEL machine from the mid-1960s), which at last achieved the objective of full isocentric rotation with the centre of rotation at a manageable height (some 120–130 cm) above the floor.

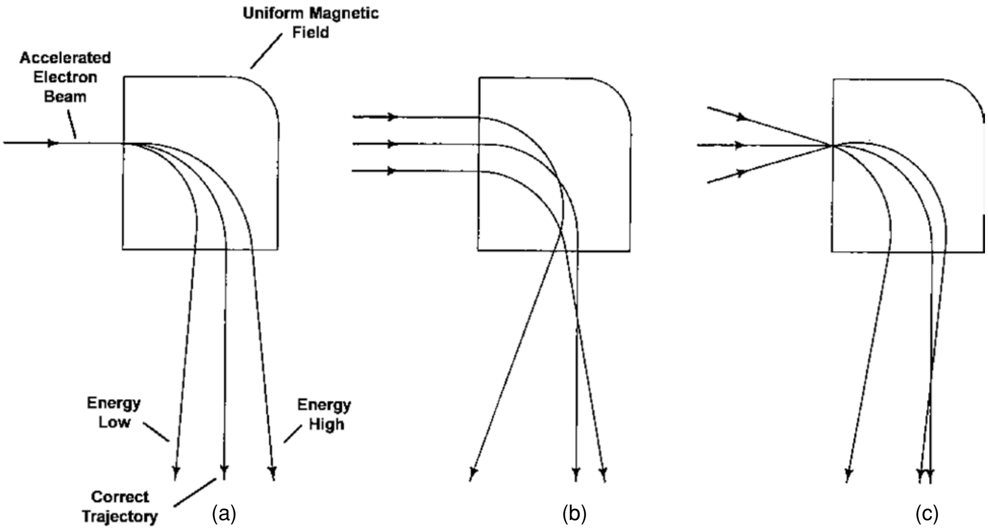

Figure 9. The problems of employing a uniform transverse magnetic field to deflect the electron beam.

The machines produced by Varian achieved this from 1962, but typically with a rather greater isocentre height.

Although the deflection of the accelerated electron beam on to the target by means of a transverse magnetic field is simple in principle, there are a number of practical problems, illustrated in figure 9. Firstly, since the passage of a beam of charged particles through a uniform magnetic field forms the basis of a spectrometer, any deviation of electron energy from the design value causes them to exit from the bending system along incorrect trajectories (figure 9(a)). However, the field flattening filter can only function correctly if the incident photon beam strikes it centrally and normally, in turn requiring the electron beam to strike the target centrally and normally. Deviations in electron energy would therefore produce a non-uniform beam profile at the treatment level. Additionally if there are deviations of input trajectory in terms of lateral or angular displacement from the correct track then (figures 9(b) and (c)) this too will result in incorrect exit trajectories and non-uniformity of the treatment beam. In the first bent beam machines these problems were addressed by monitoring the symmetry of the x-ray beam and using servo-controls to correct the machine's output whenever asymmetry was detected. However, such a system could not provide the degree of stability required for modern radiotherapy techniques.

11. Achromatic beam bending magnets

In optics a lens bringing light rays of different wavelengths to a common focus is termed 'achromatic' and this has also been used to describe a bending magnet with the ability to bring electrons with a range of energies to a common focus. To totally overcome the problem of energy spread however, the ideal magnet must bring the electrons not just to a common focus but also to a common trajectory; such magnets are termed 'doubly achromatic'. One of the first was designed by Enge (1968) and is shown schematically in figure 10. This system depends on the magnetic field strength progressively increasing in the direction shown, such that all electrons emerge at a common point and on a common trajectory, even though

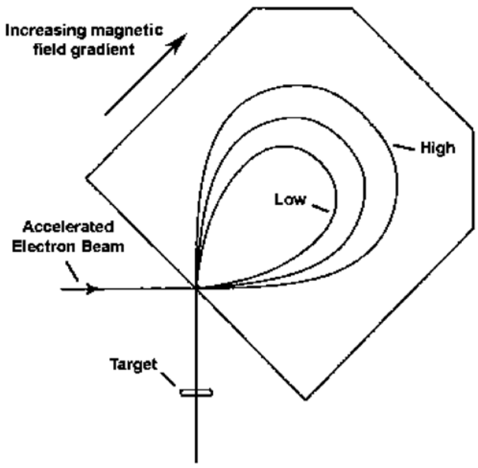

Figure 10. Enge's achromatic bending magnet.

different energy electrons follow different paths through the system. This was frequently referred to as a 'Pretzel' magnet due to the shape of the trajectories and was incorporated in the Dynaray family of linacs from Radiation Dynamics (and later from BBC/ABB). It has also been termed a 'mirror' magnet, as its action appears similar to that of a plane mirror. However this emphasizes one of the essential design weaknesses, i.e. any spatial deviation of input trajectory will be faithfully reproduced in the output trajectory. Since then, each of the major linac manufacturers has developed their own design of doubly achromatic beam bending magnet, incorporating also a degree of spatial focusing, with the result that modern bent beam machines exhibit beam stability that could only be dreamt of in the early units. A full discussion of beam bending systems can be found in Karzmark *et al* (1993).

12. Standing wave linear accelerators

Varian's first fully isocentric linac, which commenced treatments in 1962, incorporated a travelling-wave accelerator, but after Knapp *et al* (1968) had developed the side-coupled standing wave structure they adopted this approach for their low energy machines. The most significant difference with this design is to produce improved shunt impedances in excess of 100 MΩ m^{-1}, so the total length of the accelerating structure for a given energy can be greatly reduced. This enabled the design of machines which could produce 4–6 MeV from waveguides only 30–35 cm in length, so it was entirely practical to mount the waveguide vertically, in line with the target. This dispensed with the need for beam bending, and yet still producing full isocentric rotation with the isocentre at an acceptable height (figure 11). With higher energy machines however the waveguide had still to be mounted horizontally with a beam bending system, although its length is always significantly less than the equivalent travelling-wave version for a given energy.

Since the standing-wave system was first adopted it has been the system of choice for North American and Japanese manufacturers, while machines originating from the UK have opted for the travelling-wave system. There are pros and cons in respect of both, as discussed in Karzmark *et al* (1993) and Greene and Williams (1997). However both systems are still used in volume production. Of the major manufacturers Varian and Siemens use the standing-wave principle and Elekta continue with the travelling wave.

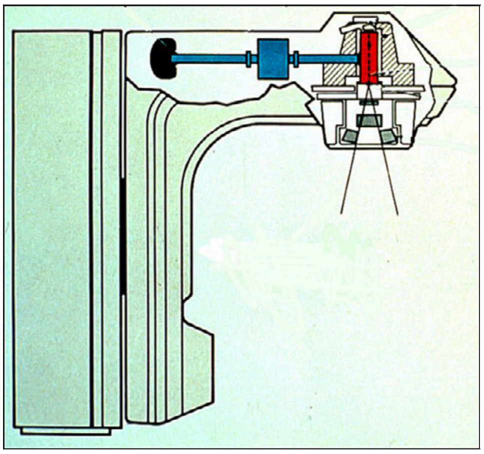

Figure 11. The 'straight-through' standing-wave configuration as used by Varian in their 4/6 MV linacs.

13. Linacs in the 1970s

By this time stable, compact, reliable and fully isocentric linacs, employing both standing-wave and travelling-wave systems, were available from a number of manufacturers worldwide and those available tended to be divided into two families. Those producing x-rays only at modest energies of 4–6 MV were often referred to as low energy, single energy or 'baby' linacs to differentiate them from those multi-modality systems offering a choice of two or three x-ray energies and multiple electron energies typically over the range 4–20 or 25 MeV.

However, it would be a mistake to think that the clinical linac was an overnight success in getting to this stage in its evolution. After initially selling a small number of machines the manufacturers experienced some lean years. For example, in 'Varian Associates—An Early History' Ginzton states 'Initially it was easy to sell a few machines because at that time there were a number of doctors ready to explore this new approach to the therapy of cancer. Then, for a number of years, we had a major dry spell . . . we had to be patient and withstand financial losses for a long time. It was a decade before it was clear that the Clinac would pay for itself. Along the way, there were many who wondered whether the idea was worth it'. By the 1970s the linac had established itself.

14. More recent linac development

Since the 1970s no further radical changes have taken place in the concepts and structures underlying the basic provision of clinical linacs. However waveguide design has continually improved as has the sophistication of beam bending systems. These factors together with other improvements such as the greatly enhanced reliability and stability of magnetrons and other components and control systems have resulted in machines with the overall stability of performance that is essential for modern conformal techniques. Additionally, modern machines have continued to develop by improving other systems and adding functionality.

In hindsight, although probably not quite so obvious at the time, there have been clear stages of development, via which the early machines have evolved into the sophisticated modern linacs we now use. These include: the initial early development from a research physics machine to a clinically viable one; basic control, safety and support systems; beam modifying and patient positioning accessories; isocentric mounting and full rotation; vacuum pumps and other vacuum component development; beam bending systems; multi-modality capability and increased energies; independent collimators and then MLC; EPIDs (fluorescent, matrix ion chamber, then amorphous silicon flat panel); computer control (then more comprehensive linac-linked information systems generally); dynamic wedges (then further dynamic control); IMRT capability and very recently readily available integral IGRT imaging systems. In parallel, the development of versatile, accurate and increasingly automated patient positioning systems has moved steadily forward from the initial basic couches or tables, as one component of a general overall improvement of geometric accuracy to meet the demands of modern techniques. Safety and control features at multiple levels of linac operation have been significantly developed, unfortunately at times as the result of radiation accidents. Trends in electronics design and modularity have been harnessed to improve stability, fault-finding and repair. Lastly and importantly, there have been significant developments in the ease of use and patient experience aspects of linac operation, in terms of system ergonomics for operators and the treatment and process environment for patients. All this evolution has been with an appropriate balancing of cost and performance to provide stable, reliable, flexible and cost-effective radiotherapy treatment machines.

It is not our intention to cover any of these in detail here, as they are discussed at length in other reviews and textbooks (e.g. Van Dyk (1999, 2005)). However, we would like to point out that many of the concepts have been around for significant periods, e.g.

- Multi-leaf collimators (MLC) had been used, in Japan in particular, since the 1960s, but were not widely taken up as the control technology then available was not easy to use. The term 'conformation therapy' was used by Takahashi *et al* (1961).
- Intensity modulation has a long history (although it was not called that at the time!) using compensators and other beam modifiers (Ellis *et al* 1959, Boyer *et al* 1999).
- Dynamic 'tracking' techniques were used to improve conformation on Co units in the 1960s at the Royal Northern Hospital, London (later the Royal Free Hospital), an extension of their earlier 1950s techniques on kV units and which were also then extended to linacs in later years (Jennings 1985).
- Film for portal imaging has a long history, of course. However Munro's 1999 review also points out that Neilsen and Jensen (1942) used a fluorescent screen to verify positioning of patients on kV therapy units, with remote real-time corrections to position, i.e. 'dynamic conformal radiation therapy—with image-guided feedback' (Munro 1999) and also that true electronic portal imaging was reported by two groups in 1958 based on screens and TV cameras.
- Add-on imaging systems with separate kV diagnostic x-ray sources had been used since the 1960s, as we have already mentioned.
- Computer control of machine movements and set up and dynamic control of beam distributions goes back to the 1970s, e.g. Levene *et al* (1978) and Kijewski *et al* (1978).

We could go on! However, of course, one main point about these pioneering efforts is that they were not taken up widely at the time. The technology was not stable, or user-friendly, and the control applied or data obtained was not easy to handle. The revolution in many of these areas came about with the introduction of widespread computer control in the 1980s and more-so in the 1990s that made many of these things a practical proposition. The same can be said for radiation oncology clinical information and control systems, where very limited 'select and confirm' had been available based on hardware interlocks since the late 1960s, but which rapidly expanded into computerized 'record and verify' systems in the 1980s and then into wide radiation oncology information systems (Greene and Williams 1997, Podgorsak *et al* 1999, Brooks 1999). We can also play this game to things we have already referenced as 1950s. For example, isocentric mounting (Howard-Flanders and Newbery 1950) was first mooted as desirable in 1906 by Kohl in a German patent (Karzmark and Pering 1973); further evidence that nothing is in isolation and everything builds on what has gone before and therefore how important it is to be familiar with the history!

15. Safety

The development of the clinical linac has its darker side with a not insignificant accident record. Of those attributable to equipment malfunction rather than human error, some of the most severe have been those arising from the ability of dual-mode linacs to produce dangerously high dose rates in the electron mode of operation under fault conditions. An early significant accident of this type occurred at the Hammersmith Hospital in 1966, when a number of patients received gross electron beam overdosage, as a result of the failure of a single component in the linac's dose control system (Lancet 1966). In response to this incident the UK Ministry of Health set up a committee to advise on how the design of control systems and methods of operation could be improved to minimize the chances of similar occurrences. The committee's title was The Radiotherapy Apparatus Safety Measures Panel (RASMP) and their recommendations were known as the RASMP 68 Regulations (DHSS 1968). These immediately became mandatory in the UK for all newly installed linacs and formed the basis of other national and international recommendations. It is salutary to note that other incidents have occurred later than that due to electron beam overdoses, albeit from different specific causes; as a few examples we can note the Zaragoza incident, the Therac 25 malfunction and a recent Polish linac misadministration (IAEA 2000).

In the 1970s the International Electrotechnical Commission (IEC), after the production of the first 'General Standard for the Safety of Electro-Medical Equipment' embarked on a programme of production of particular standards for specialized items of electro-medical equipment, and a high priority on this list was the medical accelerator. The first particular standard (IEC 1992) appeared in 1981 and covered all aspects of safety, i.e. mechanical, electrical and radiation hazards. Although medical accelerators present significant mechanical and other hazards, not surprisingly the greatest attention was devoted to radiation safety and most of this to the radiation safety of the patient. As the UK had already produced a thorough scheme of safety measures in the 1968 RASMP recommendations, most of these were incorporated into the much broader IEC document.

Although the particular standard addresses safety matters, it was widely recognized that one of the greatest safety concerns is that the patient receives the dose prescribed, which is of course determined by the integrity of a machine's performance. There have been a range of sets of national and international recommendations on machine specification, performance criteria and quality assurance from many professional bodies, e.g. in the UK (HPA/IPEM), USA (AAPM), France (SFPH), etc The IEC has also published documents (1989a, 1989b) which established a series of standard tests for every detail of a treatment machine's performance and which then recommended acceptable tolerances for the test results.

16. The future?

The immediate future will be a continuation, consolidation and further development of IMRT (Bortfeld 2006, Mackie 2006, Webb 2006) with less-widespread approaches also being thoroughly tested, e.g. IMAT, and a rapid increase in the number of installed systems having fully integrated IGRT capability, based on MV CT or cone-beam kV CT. IGRT is likely to widen to encompass other approaches, e.g. the recent papers on the feasibility of MRI and linac combined (Raaymakers *et al* 2004, Raaijmakers *et al* 2005). Of course, imaging generally is becoming more and more integrated and intrinsic to the radiotherapy process and this will continue. There is the possibility of other accelerators, such as laser-particle systems; however we feel that such major changes in the accelerating technology are unlikely to be rapidly introduced unless they show some very significant cost or other advantages. More likely will be further advances in mounting and delivery, following on from tomotherapy (Mackie 2006) and robotic arm mounted systems, more advances in beam shaping to further improve resolution and flexibility and further use of computer control, robotics and imaging to make the whole process more automated (machine, couch and patient set-up parameters; automatic control and corrections, based on image guidance and in 4D, so coping with motions from breathing, etc and also requiring less attention to initial patient position). Image guidance will increase demands for integrated and fast imaging, treatment re-planning and hence adaptive modifications both to dosimetry and positioning before delivery on a frequent basis, meaning that totally integrated automated systems will be required at the point of delivery. The discussion on open systems might be likely to recede, as the requirements on the integration of different steps in the process increase and unified solutions are seen as easier to handle this and to handle the continual development of individual process components. However the integration of information transfer between a host of imaging systems, including those on linacs, will be further demanded and we still hope to see the day when connectibility between such systems from different manufacturers does work first time! From all of this, QA demands will increase, because of the increased automation and increased complexity and also due to further increased demands on performance. Past experience from radiation accidents emphasizes that very clear approaches to safety testing and verification of new approaches are mandatory, especially as complexity increases. There will be increasing societal pressure to evaluate the cost-effectiveness of new medical technology. Whatever direction developments take, we can be certain that the pace of change will continue to accelerate.

Back to the future? Whether in science, medicine, politics, or any other area of human endeavour, we can better understand where we might be going and take more informed decisions by looking back at the past and understanding previous experience, including previous mistakes. We hope this review might have reminded the reader of some of the excitement of that early revolution in radiotherapy, when megavoltage treatment was just becoming a reality, at a time when the speciality is beginning another—image-based—revolution.

Acknowledgments

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