Cobalt-60: An Old Modality, A Renewed Challenge

Jake Van Dyk and Jerry J. Battista

Physics Department

London Regional Cancer Centre

London, Ontario, Canada

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Correspondence to: Jake Van Dyk, Head, Clinical Physics,
London Regional Cancer Centre,
790 Commissioners Road East,
London, Ontario, N6A 4L6 Canada.

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"Inconsistencies of opinion, arising from changes of circumstances, are often justifiable."

Daniel Webster, 1846

1. INTRODUCTION

The discovery of x-rays and radioactivity 100 years ago has led to revolutionary advances in diagnosis and therapy. However, it was not until the middle of the twentieth century that megavoltage photon energies became available through the use of betatrons, cobalt-60 gamma rays and linear accelerators (linacs). The increased photon penetration and skin sparing provided radiation oncologists with new opportunities for optimizing patient treatments. In recent years, several reports have considered various issues which define the "optimum" photon energy for the treatment of malignant disease[10,26,41,44]. In many of these articles[10,26], cobalt-60 is mentioned although it is generally not recommended for radiation therapy departments in the western world. Indeed, many now consider cobalt-60 as an old modality that is only useful for palliative treatments in a large department or for developing countries[36,58] with limited technical resources.

The paper by Glasgow *et al.*[16] published in this issue of *Current Oncology* reviews the use and dosimetry of a new, extended distance cobalt-60 therapy machine. The authors not only provide detailed physical considerations of this new unit but they also provide a brief comparison of the clinical use of cobalt-60 *versus* x-rays produced by accelerators. In this commentary, we extend this discussion further. We briefly review the arguments that have been presented both for and against the use of cobalt-60 as well as add some up-to-date insights and perspectives. Undoubtedly, we will not resolve this debate for all clinical situations. However, we hope that by putting "*all* the cards on the table", the cobalt-60 option will be viewed from a fairer perspective than we have seen in recent years of rapidly advancing accelerator technology. Furthermore, we also make some recommendations for the designers of cobalt-60 technology so that modernized units can be made more attractive for today's radiation therapy facility.

2. LINEAR ACCELERATORS VERSUS COBALT-60: ISSUES FOR COMPARISON

Table 1 summarizes issues for comparison for the use of cobalt-60 *versus* linear accelerators. These issues are not listed in any order of priority although they are broadly categorized according to radiation beam characteristics, machine characteristics, technical support, safety considerations and, finally, cost considerations. Different levels of importance can be assigned to each of these factors according to the local practice of radiation oncology in a cancer centre.

2.1 Radiation Beam Characteristics

2.1.1 Beam edge sharpness (penumbra)

(a) Issues for Consideration

One of the strongest arguments against cobalt-60 has been the unsharpness of the beam edge or its large penumbra. This is generally manifested by the distance between the 80% to 20% or the 90 to 10% doses at the edge of the beam. Sample data have been published by various authors[16,26,41,46]. It is important to note that there are sizeable differences between penumbras as published in the literature. These are strongly dependent on both the depth of measurement in water as well as dosimeter type and size. It is clear that for cobalt-60, the penumbra widths increase with source diameter (e.g. 1.0 to 2.0 cm), the distance between the source and the bottom of the field definer, and the distance between the field definer and the patient. The x-ray beams from linacs, on the other hand, offer penumbras which are only mildly dependent on geometry due to the small source focal spots (e.g. 0.1 to 0.3 cm)[38]. However, with increasing x-ray energies, the beam edge is blurred by more energetic electrons scattered in tissue over a greater lateral range. The effective penumbra achieved in the patient is thus significantly enlarged compared with a pure geometric penumbra, and it cannot be reduced by machine design.

There are at least four other major considerations that should be incorporated in the penumbra criterion for comparison although little quantitative data exist for these considerations. The first has to do with the radiation oncologist's ability to define target volumes accurately or consistently. The need for very precisely defined field edges is based on the assumption that target volumes and normal tissues can be defined with a high degree of accuracy. For some normal tissues and with appropriate imaging data, this assumption can be valid. However, for the accuracy of definition of planning target volumes, very little data exist. A recent study by Leunens *et al.*[27], comparing the variability of 12 physicians defining target volumes of 5 different patients with brain tumours, indicated that the estimated tumour and target sizes varied by factors of 1.3-2.6 and 1.3-2.1, respectively. Maximum variations were of the order of 1.1 to 2.7 cm in the cranio-caudad direction and 1.4-2.1 cm in the fronto-occipital direction.

The second consideration has to do with patient motion. Our desire for a tight penumbra is reasonable only if a narrow treatment penumbra can be maintained in clinical practice. However, in reality, patients undergo 20 to 30 fractions for radical therapy and usually 5 to 10 fractions for palliative therapy. As a result, the sharpness of the dose delivered to the patient is strongly dependent on the reproducibility of the patient setup relative to the beam[18,25,57]. With the recent developments in portal imaging, it is now well recognized that setup reproducibility is typically of the order of 0.5 cm for head and neck patients (up to 0.7 cm in Michalski *et al.*[35]) while it is about 1.0 cm for pelvic and thoracic treatments (up to 1.4 cm and 1.2 cm, respectively in the data of Michalski *et al.*[35]). The impact of this will be to blur the edge of the beam with respect to tissue elements near the beam edge. The net result is that even a "perfect" penumbra (i.e., a step function with "0" penumbra width) will be smeared out by beam placement uncertainty.

The third consideration has to do with organ motion. Various authors have shown that prostatic motion of up to 3 cm can occur mostly in the anterior and/or superior direction[8,25,48]. Similarly, bladder treatments involve large changes in bladder and rectal diameters[52]. Thoracic studies have shown substantial tumour movement (−1.5 cm) as a result of cardiac and respiratory cycles[45]. Similarly, in head and neck treatments, gross tumour volumes can change during a 6 week course of treatment. The net result is that our ability to reposition the involved tissues is severely limited by both a "moving target" within the patient as well as our ability to reposition the patient from day to day.

A fourth consideration takes a different perspective and has to do with the biological response of the irradiated tissues. The response of the irradiated cells generally have a sigmoidal dose-response relationship. Tumours and normal tissues behave similarly although with different dose sensitivity and slopes to the curves. These dose-response curves can be characterized by the slope at the 50% response level {often parameterized[4] by the contrast figure, γ_{50} = (%change in response) / (%change in dose)}. While there are large variations in gamma, dependent on tumour or normal tissue type[4,49], it is not unreasonable to produce a 10% change in response by a 5% change in dose, i.e., $\gamma_{50} = 2$. The net effect of this is that even a large geometric penumbra of 1.6 cm as might be found on a conventional cobalt-60 machine, could have a "biological penumbra" that is twice as steep as the physical penumbra[54]. Of course, these biological considerations are much more complex because they depend on the dose level (i.e., in which portion of the dose-response curve) and they could involve partial volume effects for the tumour and normal tissue compartments.

In summary, our usual simple preference for sharp physical penumbras is intuitive but it should be extended to consider the reality of non-reproducible patient setups as well as the biological considerations which greatly accentuate the biological penumbra.

(b) Opportunities for Improvement

While the above considerations question the need for ultra-sharp physical penumbras, there are still opportunities for improving the beam sharpness for cobalt-60 since the penumbra on a cobalt-60 machine is primarily dependent on geometric considerations. Both source size and source-to-collimator distances are adjustable parameters. A redesigned modern cobalt unit could incorporate multileaf collimators and dynamic wedges thus minimizing the need for trays for ancillary devices and thereby allowing a larger distance between the source and the field defining apparatus. Furthermore, is it not possible to redesign source capsules such that similar effective activities can be contained within smaller source diameters? We recognize the simplicity of our comments. However, there has been very little effort on improving the design of cobalt-60 units since their original design[20-22].

2.1.2 Beam Penetration (Energy)

(a) Issues for Consideration

The benefits of an increase in energy have been well documented[10,15,26,41]. Often this is reviewed from the point of view of depth dose fall off for single fields, or comparing the ratio of the dose at the depth of maximum dose to the midplane dose for parallel-opposed fields[10,41,51,55], or by considering the integral dose for typical multiple field treatments[10,26]. However, other factors must be incorporated into this discussion as well. For example, when there are superficial nodes to be treated as occur in head and neck[7] or Hodgkins disease[51], or superficial target volumes as in breast cancer patients or in total body irradiation for bone marrow transplants[56], then it is important to consider also the build-up depth for a parallel opposed pair of fields. For example, for cobalt-60 (10x10 cm^2 field, patient thickness 25 cm), the 95 % depth occurs at about 0.3 cm, for 6 MV x-rays at 0.7 cm and 25 MV x-rays at 1.8 cm[51]. These data suggest that the choice of energy is strongly dependent on the "shallowness" of the target volume relative to the skin surface. Simple generalizations based only on deeper target volumes could lead to inappropriate conclusions. Laughlin *et al.*[26] produced a figure indicating their best estimate of optimum choice of energy *versus* treatment site. However, Suit[47] in an editorial on their paper indicated that appropriately fitted cobalt-60 units could be "fully acceptable in the treatment of a large majority of the patients undergoing radiation treatment for carcinoma of the head-neck region, breast, and sarcomas of soft tissues of the extremities".

The trend today is toward conformal therapy with segmented or moving field techniques. Generally, this requires multifield irradiation or the use of arc/rotation therapy. As the number of fields increases, the advantage of higher energies over cobalt-60 radiation decreases as manifested in the dose distributions and in integral doses. A simple calculation of dose at the depth of dose maximum compared to the isocentric dose for an increasing number of fields illustrates this issue.

Table 2 shows a 186% difference in these doses (relative to the prescribed dose at the isocentre) for a single field technique when comparing cobalt-60 and 18 MV x-rays but demonstrates only a 7% difference when the number of fields is increased to 20.

(b) Opportunities for Improvement

Today there is a tremendous amount of technological development in making conformal therapy techniques aimed at linear accelerators[5,6]. On a lesser scale of developmental activity, related technology has also been implemented using a cobalt-60 tracking unit[11]. However, this technology has never been adequately commercialized to make it readily available. Cobalt-60 units could be enhanced with the application of multileaf collimators and moving field hardware/software to provide dose distributions that would be very comparable to those provided with higher energy radiations.

2.1.3 Scattering conditions/dose uniformity

(a) Issues for Consideration

While rectangular fields generally provide reasonable dose uniformity, fields with a large amount of shielding will have altered photon scattering conditions resulting in greater dose variation throughout the volume as a result of photon scatter[51]. Generally, an increase in photon energy will result in more forwardly directed scatter, yielding a more uniform dose distribution within a shaped field. Accelerators generally provide a more uniform field flatness in comparison to cobalt-60 machines, and uniformity is less prone to changes in scattering conditions.

(b) Opportunities for Improvement

First, the field flatness for cobalt-60 machines could be improved by the incorporation of flattening filters[28]. Secondly, complex irregular fields could have their dose uniformity improved by the additional use of dose compensators[29]. Such compensation is complex but can achievable with a three dimensional dose computation system.

2.1.4 Contour/inhomogeneity corrections

Under conditions of electron equilibrium, the magnitude of both contour and inhomogeneity corrections decreases with increasing energy[51]. Thus, from this perspective, higher energies are advantageous since the beams are less affected by tissue density and air gap. However, with increasing energies above 10 MV, issues related to electron transport and disequilibrium must also be considered. It is now well recognized that inhomogeneity corrections for the higher energy photon beams in low density, lung-type media are strongly affected by the lack of electron equilibrium[2,24,32,41,60]. These effects are not computed accurately on most commercial treatment planning computers. Often, for small fields and low density tissues, where an increase in dose is predicted, the effects of electron transport actually result in a decrease in dose[32]. Furthermore, this effect manifests itself at the edge of any field with an increase in physical penumbra. This was quantified by Ekstrand *et al.*[12] who showed that the ratio of penumbral width in lung to that in water magnifies from about 1.0 with 4 MV x-rays to about 2.5 with 18 MV x-rays. Indeed, some institutions[24,60] limit thoracic treatments to machines with less than 10 MV x-rays to minimize the perturbation effects of the electron disequilibrium.

Interface effects are directly related to the above discussion on inhomogeneity corrections. These interface effects are manifested at the edge of small air cavities[13,14,39], at bone-tissue interfaces, and at the interfaces of metallic prostheses as might occur in mandible reconstructions or hip prostheses. Generally, in these situations, cobalt is the preferred choice of energy since the volume of tissue under-dosed or over-dosed is smaller with cobalt than it is with the higher energies[13,14,19].

2.1.5 Dose to bone

The dose to bone compared to the dose to soft tissue is often given by f_{bone}/f_{tissue}. For the higher energies, an average stopping power ratio of tissue to bone is further incorporated into the numerator[26]. Usually, these values have been quoted for the primary beam photon spectrum[26]. However, Cunningham *et al.*[9] have shown that photon spectra change with depth and field size in the patient due to multiple scattering of photons. While conventionally, the dose to bone relative to the dose to tissue is thought to increase from 1.03 in cobalt-60 to 1.07 for 18 MV primary x-rays[26], Rawlinson[41] showed that for a 20x20 cm^2 field at a depth of 10 cm, the corresponding values are 1.08 to 1.07, respectively. Thus, there is no significant difference in dose to bone relative to dose to tissue when comparing cobalt-60 to higher energy x-rays at depth for conventional field sizes. For very large fields, as encountered with total or half body photon irradiation, the increase in multiple scatter for cobalt-60 could result in a substantial increase (−10%) in the dose to bone compared to higher energy x-rays. For total body irradiation, where irradiation of blood forming tissues is intended, the use of cobalt-60 could, indeed, accentuate the dose to bone and serve as an advantage[55,56].

2.2 Machine Characteristics

2.2.1 Dose Rate

(a) Issues for consideration

The dose rate "in air" at the isocentre of a cobalt unit depends on the source activity and the distance to the isocentre. With a half-life of 5.26 years, the dose rate also decays slowly with time (approximately 1 % per month). Maximum initial activities of 13 kCuries (481 TBq) are currently produced in nuclear reactors for cobalt therapy sources. With the standard source capsules and for a source-axis-distance (SAD) of 100 cm, a dose rate of approximately 200 cGy/min is obtained for the modern cobalt unit[16]. By comparison, the dose rate from a low energy (4 MV) linac is typically 250 cGy/min while higher energy accelerators often operate at 400 cGy/min. High energy accelerator dose rates are usually limited by safety considerations rather than the maximum electron beam current on the x-ray target. It should be noted that during a treatment, the cobalt machine dose rate is essentially continuous and constant, while the accelerator dose rate is pulsed and adjusted electronically to deliver a set dose. The pulsed nature of the accelerator beams does not seem be of radiobiological significance although it is of concern for accurate ionization chamber dosimetry due to the possibility of ion recombination.

A major advantage of an effectively constant output as found on a cobalt-60 machine is that it reduces some of the uncertainties associated with the delivery of a specified dosage to the patient. Linacs have uncertainties associated with the reproducibility of the monitor ionization chamber, field flatness, and possibly a change of energy due to a drift in the electronics. The precision of the corresponding parameters on cobalt machines is much tighter than it is on linacs. A major benefit of having at least one cobalt-60 machine in a radiation therapy facility is that it provides a means of checking radiation detectors for reproducibility in their calibrations.

Practically, the dose rate and daily dose prescription determine the daily treatment time per patient. Of course, the dose rate achieved "in tumour" depends on the overlying thickness of tissue and the beam energy. Generally, the cobalt unit is at a disadvantage with respect to dose rate achieved at the target volume. For example, as the calibration dose rate is reduced from 400 cGy/min to 200 cGy/min, the treatment time is prolonged from 0.6 to 1.4 minutes for a 200 cGy dose fraction. This simplified example assumes an overlying tissue of 10 cm, a single field size of 10 x 10 cm^2 and a linac beam of 10 MV X-rays.

The "beam on" time is only a small portion of the overall time allocated per patient treatment (typically 15 minutes) but it can impact the overall patient throughput per day by approximately 5%. Assuming an average of 40 patients treated per day, a doubling in dose rate potentially results in a gain of 32 minutes per day or the equivalent 2 extra dose fractions. With the increasing pressure on greater utilization of radiotherapy machines, a higher dose rate is clearly desirable. However, this assumes that the higher dose rate can be achieved without a loss in machine reliability or the need for additional time for quality assurance. (See section 2.3 below.)

(b) Opportunity for improvement

For cobalt-60, a higher dose rate could be achieved by increasing the source activity, by innovative redesign of source encapsulation which optimizes source packing and reduces self attenuation, or by reducing the SAD (e.g. back to 80 or 90 cm). The reduction of treatment distance from 100 to 80 cm has significant impact since it increases the dose rate by 56% ! However, this is at the expense of a reduced clearance between the patient and the collimation system. Thus, a revised design must achieve a compromise between an improved collimation system providing acceptable penumbras and acceptable dose rates while at the same time allowing sufficient clearance between the head of the machine and the patient to ensure adequate setups for most techniques.

2.2.2 Patient to Collimator Distance

(a) Issues for consideration

Practically, a greater patient-collimator distance simplifies the set up of the patient for treatment, particularly when beam modifiers are appended (e.g wedges, blocks, compensators). Dosimetrically, a reduction in this distance improves the penumbral width, assuming a constant source-collimator distance. (see 2.1.1 above). This distance is therefore based on achieving a compromise between ease of patient setup when beam modifiers are used and penumbral width.

(b) Opportunity for improvement

The clearance between the patient and collimator could be improved for a cobalt machine by the use of multi-leaf collimator (MLC) technology. For cobalt-60 radiation, the collimator leaves could be significantly thinner (1-2 cm of lead equivalent thickness) than for higher energy x-rays, and dynamic control could permit wedge or compensated fields without appended hardware. However, this should not be achieved with a loss in overall reliability of the machine. (See section 2.3 below.)

2.2.3 Isocentre Height

(a) Issues for consideration

The height of the isocentre above the floor has generally been lower for traditional cobalt units (114 cm) and this has eased the transfer of patients onto the treatment table, particularly for patients who are less mobile. On the modern cobalt unit, with a SAD of 100 cm, this height is elevated to 132 cm, approaching or surpassing the isocentric heights available on accelerators and results in difficulty in patient setup by the therapists.

(b) Opportunity for improvement

The major determinant of the isocentric height is the bulkhead of the rotating gantry which must clear the floor when the gantry is pointing upward. Thus, a more compact head structure provides opportunity for a lower isocentre. There are a number of major electromagnetic components within the head of an accelerator and it is difficult to visualize how these could be placed into a more compact structure. However, for a cobalt unit, the major bulk of the head is due to shielding which could be optimized, particularly on the "back" side of the source. If additional higher density materials were used as a back attenuator just above the source drawer, the head could be slightly reduced in height and provide additional floor clearance to achieve a lower isocentric height.

2.2.4 Radioactive Source

From a radiation physics viewpoint, the gamma rays emitted from cobalt are nearly mono-energetic compared with the polyenergetic x-rays from an accelerator. This simplifies the characterization of the source energy and thus simplifies dose calibrations and calculations. For example, the primary beam is uniformly attenuated through absorbers such as wedges. In contrast, an x-ray beam spectrum is differentially attenuated across a wedge as the beam quality changes. This x-ray beam "hardening or softening" complicates dose computations and can reduce their accuracy dependent on the dose algorithm employed[34].

2.3 Service/Maintenance Issues

Practical experience at the Princess Margaret Hospital has shown that the average down time for a cobalt unit is less than 1% while the down time for linear accelerators increases with increasing complexity from about 3% for a single low energy unit to about 11% for the 25 MV type machine[41]. Similar data was quoted by Das and Kase[10] who observed a 3% down time for a 4 MV machine compared to about 5-7% for higher energy units. Our own data, at the London Regional Cancer Centre, demonstrate similar trends although with lower down times during clinical hours. For the higher energy machines, the clinical down time is about 3%, for single low energy linacs it is about

2%, and the cobalt unit has a down time that is less than 0.4%. However, to consider overall costs, the after hours preventative maintenance time should be added. In our centre, this corresponds to another 3% and 1% of clinical hours for linacs and cobalt, respectively. Of course, the expertise of maintenance staff required is directly related to the complexity of the treatment machines and this affects the maintenance costs accordingly.

2.4 Safety Considerations

2.4.1 Radiation protection

From an ecological viewpoint, radioactive sources pose an environmental hazard while being transported, while in service, and finally at disposal time. Furthermore, the source is always radiating whereas the linac x-ray beam is switchable to the "off " state at will. In practice, the cobalt source is shielded by a bulky head assembly which attenuates the continuous stream of emitted gamma rays. The advantage is a constant dose rate, but at the risk of a radiation accident as might occur if the source "gets stuck" in the "on" position. Clearly, the production of x-rays is more controllable, although there have been serious accidents due to inadequate linac controls[1,23,30]. With the current improvements in hardware and a software, the risk from linacs now appears to be smaller. The issue of source disposal is of substantial concern if inadequate legislation is in place as has been demonstrated by improper disposal procedures in Mexico and Brazil. The solution for this, however, is not the avoidance of cobalt but rather the development and implementation of appropriate regulations for the disposal of cobalt sources.

An additional hazard of the higher energy linacs (>10MV) is the production of neutrons as a result of photonuclear reactions in the head of the machine. For such machines, the greatest exposure to the radiation therapists operating the machines is due to residual radioactivity from both the machine and treatment room walls[40].

2.4.2 Pacemaker concerns

An increased number of patients are seen in radiation therapy departments with implanted pacemakers[31,33] as a result of an aging population combined with increased indications for the insertion of permanent pacemakers. Pacemaker faults are potentially generated either by interference due to electromagnetic radiation or by ionizing radiation. The level of concern for both of these is controversial. Conservative recommendations by an AAPM Task Group[33] suggest close monitoring of patients treated on linear accelerators to observe any potential effects of electromagnetic interference and maintaining total dose levels to pacemakers to less than 2 Gy. In England, where no formal recommendations regarding the use of pacemakers have been adopted, a survey indicated that about one-half of the radiation therapy departments treated their patients with pacemakers on cobalt units in preference to linear accelerators[31].

2.5 Cost Considerations

Cost analyses have been performed by various authors[10,41,53]. Rawlinson[41] has done an analysis in which he compared the annual operating costs for a cobalt unit, a low energy linac and a high energy linac including estimates of capital depreciation of the machine and the building as well as maintenance costs. The results indicate an annual cost in 1986 Canadian dollars of $38,100, $122,800 and $181,800 for the operation of a cobalt machine, low energy linac and a high energy linac, respectively. This indicates that a low energy linac costs more than 3 times as much as a cobalt unit and a high energy linac costs more than 5 times as much to operate. As indicated above, patient throughput due to a lower dose rate on cobalt differs only by about 5 to 10%. Clearly, there is a substantially reduced operating cost for treating patients on cobalt-60 machines.

3. DISCUSSION

The treatment of malignant disease by radiation therapy is performed by a large range of radiation therapy technologies including cobalt-60 machines and high energy linacs combined with a gamut of ancillary devices many of which are computer-controlled. Much of this advanced technology is purchased and implemented with relatively little consideration for actual cost/benefit but an assumption that higher energies, smaller penumbras, and higher dose rates "obviously" provide improved therapy. While there are indications that institutions using advanced technologies have better patient outcomes[17], it has never been proven that this was directly due to the machine energy levels and could well be due to surrogate issues related to staff quality that is associated with institutions having more sophisticated equipment.

Recently, in times of economic constraint, much more consideration needs to be given to cost/benefit of radiation therapy[3,37,42,43,59]. A maxim in today's culture of reducing costs is that we need to provide the most effective therapy at the lowest possible cost. This commentary has attempted to provide a comparison of cost and benefit issues associated with cobalt-60 and megavoltage x-ray utilization. This discussion is summarized in Table 3 which contains a synopsis of the use of cobalt-60 versus low, medium, and high energy x-rays combined with a list of opportunities for change. It is clear from this Table that there is still a role for cobalt-60 radiation. However, this role needs to be defined in the context of individual radiation therapy institutions. Clearly, a small institution cannot purchase the entire range of radiation therapy equipment and will have to chose the minimal number of therapy units while providing maximum overall baseline service to its patients. For such institutions, dual energy linacs with one low and one medium to high energy x-ray mode along with electron capabilities will provide the maximum flexibility. Larger institutions, however, will require more therapy units and, therefore, will have the flexibility to purchase a greater variety of equipment.

A review of Table 3 indicates that cobalt-60 can still play a significant role in treating those patients whose target volumes lie near the surface, or those anatomical regions where the patient separation is small, or those target volumes that can be treated with multiple conformal therapy beams. If we make some simple assumptions about capabilities and case mix then we can provide a crude estimate of the percentage of patients that could be treated by cobalt-60 radiation. First, about 20% of patients receiving radiation therapy are treated for cancer of the breast and another 17% are treated for cancer of the lung[50]. If we assume that about one quarter of the breast patients have chest wall separations that are smaller than 18 cm and one third of the lung patients could be treated by cobalt-60 then this already represents 11% of all the patients treated. Add to that one third of the head and neck patients, three quarters of the sarcoma patients, and one quarter of the palliative cases then we can estimate the total percentage that could be treated with cobalt-60. This is summarized in Table 4 and demonstrates that one out of every 4-5 megavoltage therapy machines could be a cobalt-60 unit. These estimates are based on today's procedures. With an improved design, cobalt-60 could possibly be used for about 30% of all patients treated with radiation therapy with a corresponding reduction in overall operating costs. However, such a benefit is only achievable in an economic and health care system that rewards minimal cost treatments and where physicians are not rewarded for the use of more complex technologies. These estimates are also based on resources available in the western world. Where resources are more constrained, as in developing countries, the fraction of patients treated by cobalt-60 can be increased substantially and will be strongly dependent on local resources.

4. CONCLUSIONS

In conclusion, cobalt-60 can still be regarded as a very viable and cost effective option for the treatment of a sizeable fraction of cancer patients ***assuming that the technology is improved***. It is our estimate that, in the western world with readily available financial resources, at least 25% of cancer patients requiring radiation therapy could be treated with this modality. This is only possible in departments that use more than two or three therapy machines since higher photon energies and the use of electrons will still be required for those patients with larger separations and deep seated tumours and those patients with tumours near the skin surface. In developing countries, where financial and technical resources are severely restrained, a much larger fraction of patients requiring treatment can benefit from cobalt-60 therapy. Recognizing its limitations, the "cobalt-60 challenge" is to redesign existing technology to offer cobalt-60 therapy as a cost/beneficial alternative for a well-defined fraction of the patients requiring radiation therapy. In times of economic restraint, it becomes even more important to strike a delicate balance between the use of accelerators and cobalt-60 machines.

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Table 1. Criteria for Comparing Radiation Therapy Machines

A. Radiation Beam Characteristics

1. Beam edge sharpness (penumbra)
2. Beam penetration (energy)
3. Scattering conditions/dose uniformity
4. Contour/inhomogeneity corrections
5. Dose to bone

B. Machine Characteristics

1. Dose rate
2. Patient collimator distance
3. Isocentre height
4. Radioactive source versus x-rays

C. Service/Maintenance Issues

D. Safety Considerations

1. Radiation protection
2. Pacemaker concerns

E. Cost Considerations

Table 2. Comparison of the dose at depth of maximum dose (in %) to the dose at the isocentre (100%) as a function of number of fields. Depth to isocentre = 20 cm (equivalent to a lateral patient thickness of 40 cm). Field size is 10x10 cm^2, SAD = 100 cm.

Number of fields	Cobalt-60	6 MV	18 MV
1	390	284	204
2	206	159	126
4	99	79	63
6	66	53	42
10	40	32	25
20	20	16	13

Table 3. An evaluation of cobalt versus linacs combined with opportunities for improvement
* low score   *** medium score   ***** high score

Feature	**Cobalt**	**Low energy linac**	**Medium energy linac**	**High energy linac**	**Opportunity for change**
A. Radiation Beam Characteristics					
1. Beam edge sharpness (penumbra)	*	****	***	**	Cobalt source redesign and use of MLC
2. Beam penetration (energy)					
Parallel pair: Small separations < 14 cm	****	*****	****	***	Development of conformal therapy techniques using moving fields combined with MLC.
Medium separations 14 - 20 cm	****				
Large separations >20 cm	*	**	****	*****	
3-4 fields: Thorax, Pelvis	*	**	***	*****	
Multiple fields, arcs, rotations	****	****	*****	****	
3. Scattering conditions/uniformity	*	**	****	****	Use of flattening filters and dose compensators.
4. (a) Contour/inhomogeneity corrections	**	***	**	**	
(b) Build-up/build-down/interface effects	****	***	**	*	
5. Dose to bone	***	***	***	***	
A1. Optimum Energy By Site (refs)					
1. Brain	****	*****	***	**	
2. Head and neck	*****	****	**	*	
3. Breast	****	*****	**	*	
4. Lung/oesophagus	**	*****	***	*	
5. Lymphoma	**	*****	***	*	

6. Pancreas	*	*	***	*****	
Table 2 (continued)					
7. Pelvis	*	**	****	*****	
8. Extremity soft tissue sarcoma	*****	****	**	*	
9. Pediatrics	****	*****	**	*	
B. Machine Characteristics					
1. Dose rate	*	****	*****	*****	Redesign source encapsulation
2. Patient collimator distance: Co-60, 80 cm Co-60, 100 cm	* ***	***	***	***	Implementation of MLC
3. Height of isocentre above floor:Co-60, 80 cm Co-60, 100 cm	**** *	**	**	**	Improved shielding to reduce head size.
4. Gamma rays vs x-rays	**	*	*	*	
5. Constancy of output	*****	*	*	*	
C. Service/Maintenance	*****	***	*	*	
D. Safety Considerations					
1. Radiation protection	***	****	**	**	
2. Pacemaker concerns	*****	***	*	*	
E. Cost Considerations	*****	***	**	*	

Table 4. Clinical sites which can benefit from the use of cobalt-60. Based on data from reference 50 as well as data from the London Regional Cancer Centre for 1992-93. (Sites where cobalt is not generally recommended are excluded.)

Clinical site	% of total patients for OCTRF[1] (LRCC[2])	Potential fraction treated with Co-60	% of total patients who can potentially be treated with Co-60 for OCTRF[1] (LRCC[2])
Brain	2.2 (6)	0.75	1.7 (4.5)
Breast	20 (18)	0.25	5.0 (4.5)
Head and neck	5 (15)	0.33	1.7 (5)
Lung/respiratory	17 (9)	0.33	5.6 (3)
Thyroid/sarcomas	2.4 (1[3])	0.75	1.8 (0.8)
Pediatrics	1.3 (1[3])	0.75	1.0 (0.8)
Palliative for other sites	20 (26)	0.25	5 (6.5)
Total	68% (76%)		22% (25%)

1. OCTRF = Ontario Cancer Treatment and Research Foundation. The OCTRF data include all new cancer cases registered by clinical site for the 1992-93 fiscal year.
2. LRCC= London Regional Cancer Centre. The LRCC data include radiation therapy patients for the 1992-93 year
3. This is an estimate since detailed data were not available.