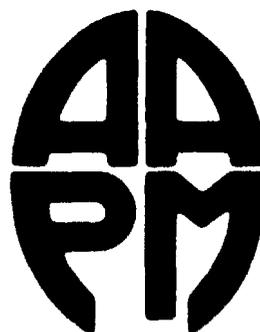


AAPM Code of Practice for Radiotherapy Accelerators



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AAPM Code of Practice for Radiotherapy Accelerators

REPORT OF AAPM RADIATION
THERAPY TASK GROUP 45

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AAPM code of practice for radiotherapy accelerators: Report of AAPM Radiation Therapy Task Group No. 45

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PART 1: INFORMATION FOR RADIATION ONCOLOGY ADMINISTRATION

The present document has been developed as an update of the 1975 AAPM "Code of practice for x-ray therapy linear accelerators,"¹ which was largely devoted to machines producing x-ray beams of 10 MV or less. It is intended to cover all medical electron accelerator equipment currently on the market, including photon as well as electron beam machines. The part 1 of this document is addressed to the radiation oncology administration which may include chief radiation oncologist, radiation oncology department administrator or a hospital/free-standing center administrator. In these guidelines, the AAPM recognizes the importance of team effort between administrators, radiation oncologists, physicists, dosimetrists, radiation therapists (formerly known as radiation therapy technologists), and engineers, required to establish an optimal radiation oncology program.

In the treatment of cancer with radiation, the radiation oncologist prescribes a treatment regimen to cure or control the disease while minimizing complications due to damage to normal tissue. In general, published clinical and experimental results show that dose response curves, for tumor control or for normal tissue damage, are variable. For some treatment sites these curves are very steep in the therapeutic dose range, i.e., a small change in the dose can result in a large change in clinical response. Moreover, the prescribed doses are often, by necessity, constrained by tolerance doses for normal tissue. Therefore, for optimum treatment, the radiation dose must be delivered accurately.

The International Commission on Radiation Units and Measurements (ICRU) has recommended that the dose delivered be within 5% of the prescribed dose.² Considering the many steps involved in delivering dose to a target volume in a patient, each step must be performed with an accuracy of much better than 5% to achieve the ICRU recommendation. For example, if only three steps were involved (e.g., tumor localization, dose calculation, and machine calibration) and assuming the uncertainties combine in quadrature, better than 3% accuracy would be required for each step.

Over the past 2 decades, great strides have been made in the technology of diagnostic imaging as a basis for tumor localization, the physics of radiation dosimetry, computer-assisted radiation treatment planning, and in the technology of external beam radiation machines, particularly electron

accelerators capable of delivering both x-ray beams and electron beams. Higher energies have become available, as well as more sophisticated beam control techniques, including automatic machine setting, treatment monitoring, and recording. Programmable multileaf collimators enabling dynamic treatment are becoming available and offer new possibilities, but also new risks and responsibilities, and thus challenge capabilities of the radiation oncologist and the physics and therapy staff.

These technological developments offer a wider spectrum of beam energies and technical capabilities, with new therapeutic possibilities. However, they pose new questions and problems to not only the radiation oncologist and the physicist, but also to the institution's management team. Decision making in regards to new radiation treatment facilities involves many steps and many expertises. It should start with the formulation of the radiation oncology needs of the institution based on the expected development of patient population and include the development of specifications for all proposed equipment, housing and support requirements, selection of the equipment itself, acceptance testing, commissioning, quality assurance, maintenance, and finally initial and continual staff training. Compliance with state and federal regulations, as well as recommendations from bodies such as the National Council on Radiation Protection and Measurements (NCRP), must be assured.

The standard of practice requires the radiation oncologist to have a precise knowledge of the expected radiation dose distribution throughout the irradiated volume of the patient prior to initiating treatments. This knowledge is based on physical information about the radiation beam, alterations caused by beam modifiers and various tissues of the body, and on how several beams might be combined. The multiple treatment modalities and energies available with modern treatment machines require sophisticated control and supervision. The increasing application of computer control poses special advantages as well as risks of a new nature, as some recent incidents have demonstrated.^{3,8} If anything, the need for a well-trained, in-house interdisciplinary staff for a modern facility has increased, and hospital administrations should not underestimate this aspect.

If the potential advantages of modern medical accelerators are to be realized, then the decision to purchase such equipment must be accompanied by the concomitant decision to have the required dosimetry and treatment planning equipment with the appropriate staffing levels of qualified radiation oncology physicist(s). In that regard, this code of practice recommends the radiation oncology physicist be certified in radiation oncology physics by either the American Board of Radiology, American Board of Medical Physics, or the Canadian College of Medical Physics (Appendix A). We recommend that radiation oncology facilities be staffed at levels that closely follow the guidelines given in the "Blue Book," the Report of the Inter-Society Council for Radiation Oncology.⁹ In particular, we recommend that facilities with even a single multimodality megavoltage medical accelerator have a full-time qualified radiation oncology physicist. A part-time consulting physicist alone does not provide radia-

tion oncology physics services of a quality necessary for state-of-the art radiation treatment.

The radiation oncology physicist is responsible for the acceptance testing, commissioning, calibration, and periodic quality assurance (QA) of the therapy equipment and directly oversees the determination of radiation dose distributions in patients undergoing treatment (i.e., computerized dosimetry planning or direct radiation measurement). The radiation oncology physicist should help define the specifications for the purchase of the treatment unit(s), therapy simulator(s), and treatment planning system. The radiation oncology physicist should be involved in the design of the facility and must survey the environs after installation of a radiation treatment machine in order to assure compliance with the applicable state and federal laws (as well as with the recommendations of such bodies as the National Council of Radiation Protection). The radiation oncology physicist shall certify that the treatment machine is performing according to specifications after it is installed, generate the data necessary for the accurate treatment planning and delivery of the radiation, and outline a written QA- procedure which includes tests to be performed, tolerances, and frequency of the tests. The responsibilities of a radiation oncology physicist are listed in Appendix B.

The newer generation multimodality (x ray and electron) medical accelerators utilize, in many instances, computer technology in their control systems. Such accelerators have the potential for massive overdoses to the patient as a result of software flaws. Therefore, the radiation oncology physicist must carefully scrutinize the control systems of such machines during the acceptance testing period, and after software updates, paying particular attention to verifying what happens when beam operating parameters are edited, e.g., switching between photon and electron modes.

Regular maintenance of a radiotherapy accelerator is required. We recommend that this must be overseen by the radiation oncology physicist. While the supervising radiation oncology physicist is not responsible for performing the actual machine maintenance, he or she is responsible for the release of the treatment machine into clinical service after maintenance and for documenting that any alteration caused by the maintenance and repair schedule does not affect the accelerator vis à vis the standards promulgated in this code of practice. However, it is stressed that the qualified radiation oncology physicist is the sole individual who can make a decision on the working conditions of a medical accelerator for patient treatments.

In-house service may be appropriate for institutions large enough to assume the associated financial risks for failure of components and failure to successfully perform service in a timely manner. Smaller institutions may be more wisely advised to invest in and negotiate service agreements with the equipment vendor or other trusted parties. Additionally, the liability of the vendor is tapped to your support by doing business in this manner.

It is necessary that the qualified radiation oncology physicist have the appropriate equipment and test instrumentation needed for beam calibration, acquisition of beam data for the treatment planning computer, and the required periodic QA

tests. A suggested list of the necessary equipment is listed in Appendix C.

The implementation of increasingly complex treatment planning systems and treatment units in radiation oncology centers and the increasingly complex treatment procedures require formal QA and preventive maintenance programs for these units. Also, the use of radiotherapy accelerators in special procedures such as intraoperative electron therapy, stereotactic radiosurgery, and conformal three-dimensional therapy require additional and even more stringent QA procedures to insure the level of confidence in the accuracy of dose delivered. It should be noted that commissioning for special procedures is not routine for all clinics and must be thoroughly discussed between the medical physicist and the radiation oncologist with an eye toward the total complement of personal and equipment available locally to successfully perform special procedures. For any new procedure it is important for all to understand the setback, which could occur both locally and globally, for promising procedures mishandled by those not yet fully aware of the intricate details of a new procedure.

In summary, the decision to provide a community with radiation oncology facilities not only involves a decision to enlist the services of a radiation oncologist and radiation therapists, purchase and install the treatment planning and treatment delivery equipment, and to arrange for its proper maintenance, but also to provide for the support of a qualified radiation oncology physicist and to insure he or she has access to the appropriate dosimetry instrumentation. Adequate support staff such as medical dosimetrists, block makers, etc. is also essential for a cost-effective operation of the physics service. Also it is stressed that proper treatment is a team effort, and making machines available for an appropriate amount of time during normal working hours encourages timely and complete quality assurance.

A few comments on terminology are in order. There are three levels of imperatives distinguished in this report. [This language was taken from a draft version of the TG-40 report and was provided to us by J. Purdy, who is a member of TG-40 (Chairman: G. Kutcher).]

- (i) Shall or must: These terms are applied when the imperative is dictated by law.
- (ii) Recommend: Phrases like "we recommend" and "requires" are intended to convey that the task group considers the procedure referred to as important. If modification is considered, we recommend that it would occur only after careful analysis demonstrates that quality would not be degraded. When a tolerance level or frequency of testing is given, it can be assumed to be a recommendation or law.
- (iii) Should: There are many aspects of QA where tolerance levels and frequencies cannot be given, and in which quality can be maintained via many different approaches. In these instances, which apply to many aspects of QA, modals like "should" are used. The task group recognizes the complexity of the treatment planning and treatment process, and the inadvisability and impossibility of giving precise direction to QA in

this respect. However, the task group considered it important to suggest avenues for such quality assurance.

The AAPM is concerned that the recommendations in this report must account for two incompatible principles, namely, that our recommendations should reflect the highest standards, and at the same time that costs will inevitably increase to meet those standards—especially when those standards approach their practical limits. We have no ready answer to this dilemma. Nevertheless, we have tried to balance these principles by reporting what we consider to be standards of practice, or where none clearly exist, to suggest standards and procedures which we feel are consistent with other standards of practice and that could be attained with costs comparable to those standards.

PART 2: A CODE OF PRACTICE FOR RADIOTHERAPY ACCELERATORS

I. INTRODUCTION

Ionizing radiation is one of the principal modalities used in the treatment of cancer. The goal of radiation treatment is always twofold: (i) to control the malignancy and (ii) to avoid unacceptable damage to normal tissues. Optimal radiation treatment requires accurate and timely diagnosis, accurate determination of the target volume and critical organs at risk, proper dose prescription, correct delivery of the prescribed treatment and diligent follow-up. Careful documentation of every phase of the radiation treatment process is essential for clinical evaluation. The treatment process itself requires a knowledge of the biological effects of radiation on body tissues and organs, and the body as a whole. It is also important to understand the physical nature of ionizing radiation, its basic interactions with matter, and the processes of energy deposition in tissues. The physics staff is therefore not only responsible for proper functioning of the radiation equipment but also for enabling the radiation oncologist and therapist to deliver the best possible radiation treatment to within a known accuracy and precision, using available resources. The planning and delivery phases of radiation treatment are the responsibility of specially trained staff including radiation oncology physicists, dosimetrists, and therapists. Understanding the technology of medical accelerators requires high-level knowledge of radiation physics and familiarity with mechanical, electrical and electronic engineering, and computer science. Thus good radiation oncology practice requires close interaction between practitioners of a spectrum of disciplines, including oncology, medical technology, and radiation physics.

The present document is the result of the AAPM's efforts to provide guidelines which enable the safe and accurate application of external radiation beams from high energy radiotherapy accelerators. It is intended as an update to the "Code of practice for x-ray therapy linear accelerators," a document published by the AAPM in 1975.¹ As indicated in its title, the contents of the original code of practice were limited to cover linear accelerators producing x rays of 10 MV or less. Since 1975, radiotherapy accelerators that are capable of pro-

ducing higher energy x-ray, as well as electron, beams have become commercially available for routine use. Many machines can produce beams with several electron energies, and some have dual x-ray energy capabilities. In addition to linear accelerators, there are now a few microtrons coming into use. Techniques using single fractions of high doses such as intraoperative electrons, stereotactic radiosurgery, etc. have emerged. More recently, there has been an evolving interest in programmable portal shaping using multileaf collimation. This development promises a new interest in conformal therapy and poses a challenge to a radiation oncology team to develop methods that ensure safe and efficacious use of these new technologies.

The broader capabilities of radiotherapy accelerators require greater expertise of all involved in their use: radiation oncologists, physicists, dosimetrists, therapists and supporting technical staff. The selection, installation, and clinical use of modern radiation equipment involves:

- (1) development of the specifications of the radiation equipment;
- (2) design and construction of the facilities to accommodate the selected radiation equipment, including radiation shielding;
- (3) installation of the selected radiation equipment;
- (4) verification of radiation safety in the environment of the radiation facility;
- (5) acceptance testing of the installed equipment;
- (6) commissioning of the accelerator for active clinical use;
- (7) training of the staff in the safe and efficacious use of the accelerator;
- (8) development and application of a comprehensive quality assurance (QA) program.

The present document is intended to apply not only to radiotherapy accelerators that produce x-ray beams but also to those that produce electron beams. The reason for including both x-ray and electron beams in one document is that in most technical, radiation safety and practical operational aspects, these machines pose very similar problems.

At present the vast majority of radiotherapy accelerators in the USA are linear electron accelerators. A modern medical accelerator is a large, heavy, and complex piece of equipment which demands careful attention to the design of the facility to house it. The various aspects of these machines, such as the type of gantry mount, pose special construction problems. In particular, because of the size and weight of the machines and the radiation shielding requirements, one must take into account the special requirements for construction of the room which contains the machine. In particular, one must pay close attention to the thickness and composition of the floor, ceiling, and walls of the room and the occupancy of adjoining rooms by hospital personnel. The power and cooling requirements, coupled with a considerable amount of computerized control requirements, necessitate special attention to the unique needs for electrical, water, and air supplies to the accelerator room.

The operation, control and quality assurance of modern radiotherapy accelerators often involves a considerable degree of computerized control and data processing techniques.

For all practical purposes, acquisition of such equipment requires making a choice between a limited number of commercially available FDA approved accelerators. Nevertheless, the radiation oncology team that is ready to purchase an accelerator is faced with the complex task of selecting the appropriate machine from those commercially available and developing the necessary specifications to meet the team's clinical needs. This process should include:

- (1) a careful study of the clinical needs;
- (2) a careful study of technical and physical specifications of the commercially available equipment, including the technical and operational characteristics of the "patient support system" and any essential accessories;
- (3) an inventory of available space, available funds;
- (4) available or needed physics or therapy staff, and available in-house technical support;
- (5) specification of the acceptance testing process;
- (6) an analysis of the financial implications, including warranties and the possible need for maintenance contracts;
- (7) a thorough investigation of the quality of available service networks in the geographic region.

This work should be performed with close cooperation between the radiation oncologists, physicists, technical staff, and administration of the institution. It should produce specifications that meet or exceed the requirements of building codes, including fire department and radiation safety regulations.

The purpose of this document is to set a standard of practice and to provide educational material for all practicing physicists. A varying depth of detail is presented for different sections. Those sections which deal with new procedures (such as stereotactic radiosurgery) and for which literature is not easily available (such as acceptance testing) are treated in more detail than others.

II. FACILITY PLANNING AND RADIATION PROTECTION

A. Shielding design

The initial work on a new radiotherapy facility begins with the design of the room. One needs to shield the areas outside the room to the levels specified by current state regulations and NCRP guidelines. For high energy machines with nominal energies greater than 15 MeV, the maximum permissible exposure should include the effects of neutrons as well as photons, with the appropriate neutron quality factors taken into account. There have been changes recently in the recommendations for maximum permissible exposure limits (see Sec. II C below) and the newly revised recommendations must be adhered to for new facilities.

The space available for construction is a very important factor in the design of the facility. If the facility is to be built in a new area, there will be few, if any, constraints on the design. On the other hand, if a machine is to be fitted into an area in an existing building or an enclosed space, especially if the new machine is to replace one of lower energy, major constraints on the design and construction could exist. For example, lead and steel instead of or in addition to concrete

may have to be used in the construction of such facilities because of space limitations. Additional shielding should be placed on the outside of the room, if at all possible.

Probably the most important question to be answered in shielding design is whether or not a maze can be installed in the room. The advantage of a maze is that a relatively light-weight door can be used at the entrance to the facility. If once-scattered- or leakage-radiation can strike the door, the door may have to be very heavy. Moreover, at high energies where neutrons have to be taken in account, borated polyethylene or its equivalent should be added because lead will have only a minor shielding effect on the neutrons. This increases both the thickness and weight of the door which, as a consequence, will be quite expensive. Such a massive door has a longer opening time and will have the effect of increasing the overall treatment times. This type of door must be of a very conservative mechanical design to ensure against failure. To avoid the possibility of losing quick access to the radiotherapy patient due to a door failure, a second mechanism for entrance into the room should be provided. A manual method for opening the door should be available at all times. Another precaution is to have the operating system of the door on the outside of the treatment room so that the opening mechanism is accessible from the outside.

The optimal location and orientation of the machine within the room takes into account constraints mentioned above. In general, it is desired that the primary beam be directed toward unoccupied or minimally occupied areas (certainly not toward the control console). This selection process is a very time consuming task if the calculations are performed by hand. Recently, computer programs enabling the user to optimize the room design have become available for large computers as well as for personal computers (pc's). (Some of these programs can be obtained from the AAPM Software Exchange Program.)

A safety feature not currently found in all radiotherapy treatment rooms but might prove useful is a "search button." The function of this device is to ensure that only the patient is present in the room while the beam is on. Its operation is as follows: after the patient has been set up, the radiation therapist, on leaving the room, presses the search button. If the door is closed within a specified time period, say 20 s, the machine can be turned on. However, if the door is opened for any reason, the machine cannot be turned on without entering the room and restarting the search procedure. The search button needs to be strategically placed to ensure that an effective search is automatically carried out during the exit of the therapist from the room.

Another useful device for reducing exposures in occupiable areas outside the room, if the entrance to the treatment door is recessed, is a photocell which will turn off the beam if anyone approaches the door. Often, high exposure rates can be measured at the door jambs and on the floor but at a distance of a few feet these levels are generally considerably lower. Some facilities have used barrier gates in order to avoid inconvenient turning off of the machine through inattentiveness.

Neutron shielding must be considered for all machines with a maximum bremsstrahlung energy of 10 MV or

greater. Neutrons are generated by the interaction of the high energy x rays with high Z materials such as lead and tungsten located in the collimator head (with photon energy thresholds as low as 5.7 MeV). The principal sources of neutrons are the target, the primary collimator, the flattening filter, and the movable photon jaws. The neutron energy spectrum is similar to a fission spectrum, with a most probable energy of about 1 MeV. Such neutrons are not easily absorbed by high Z materials. Concrete, because of its high water content, is the most practical and least expensive shielding material for neutrons. Where concrete is not appropriate (e.g., in doors) borated polyethylene should be substituted. In general, the neutron fluence in the primary x-ray beam is approximately 2-3 times greater than the neutron fluence outside the beam in the treatment plane. If concrete barriers are designed to protect against photons, it may be that these barriers will also provide adequate protection against neutrons (although this should be verified by calculation). However, if lead or steel is used for shielding one must add either concrete or borated polyethylene to allow for proper neutron shielding. In the case of doors where photon shielding materials are limited to lead or steel one must include borated polyethylene or an equivalent neutron shielding material on the inside of the door preceding the lead or steel on the outside. As noted above, this is a case where a maze near the door becomes important in the room construction.

Simple formulas for the calculation of exposure at the end of a maze can be found in NCRP Report #51 for x rays" and NCRP Report #79 for neutrons." For details of shielding air ducts and electrical conduits, see NCRP Report #51.¹¹ A recent review of shielding design considerations is available in the 1990 AAPM summer school proceedings.¹²

B. Radiation protection survey

After the installation of a medical accelerator the radiation levels outside the room in all accessible directions must be measured using appropriate radiation survey meters. The methodology of a radiation protection survey for photons is fairly standard and has been discussed at length previously. A good reference for this purpose is NCRP Report #51.¹¹

For high energy machines with a nominal energy above 15 MeV, measurements should also be made of neutron leakage both inside and outside the room. A description of the source of these neutrons is given in NCRP Report #79.¹² Methods for measuring neutron fluence and absorbed dose rates may be found in the same publication and also in AAPM Report #19.¹⁴ In general, an activation technique using phosphorous is recommended for measurements in the primary beam with peak bremsstrahlung energy above 20 MV. For energies below 20 MV and inside the room the phosphorous and moderated foil technique using either indium or gold can be employed. The recent development of superheated drop detectors for neutrons with low sensitivity to photons makes them an attractive alternative to the previous methods.¹⁵⁻¹⁸

C. Regulations on radiation protection

There are two regulatory and advisory bodies that deal with radiotherapy equipment. Cobalt-60 teletherapy units come under the jurisdiction of the Nuclear Regulatory Commission (NRC). Safety aspects of linear accelerators are governed by the NCRP and by individual states (Suggested State Regulations for Control of Radiation, SSRCR). The state's requirements are (in almost all instances) the same as those recommended by the NCRP. However, there is a certain amount of overlap between the various sets of regulations and some regulations adopted by the NRC are applicable to linear accelerators, as will be made clear below. Also, there is a lag time until states adopt NCRP changes.

Recently, the NCRP (Report #91)¹⁹ has revised its recommendations on the exposure limits to ionizing radiation for radiation workers and the general public. As before, the effective dose equivalent limit for occupational exposure is still 50 mSv/yr (5000 mrem/yr). However, two changes affect shielding design. First, the annual effective dose equivalent limit for continuous or frequent exposure for the general public is now 1 mSv (100 mrem)/yr or 0.02 mSv (2 mrem)/wk. This is a factor of 5 lower than required by prior standards. The second change is that the quality factor for neutrons has been increased from 10 to 20. This implies that, for neutrons, the recommended limits for frequent public exposure have been reduced by a factor of $5 \times 2 = 10$. Thus, for a high energy accelerator one has to plan for a maximum dose equivalent for x rays plus neutrons of 0.01 mSv (1 mrem)/wk in areas occupied by the public. One of the regulations that had been used in the previous design criteria for linear accelerator facilities was the maximum permissible dose equivalent of 0.02 mSv (2 mrem) in any 1 h, which was tied in with the 0.1 mSv (10 mrem)/wk limit for the general public. Now that the maximum permissible dose equivalent for the public is 0.02 mSv (2 mrem)/wk, the old rule of less than 0.02 mSv (2 mrem) in any 1 h effectively has no relevance now.

It should be noted that the shielding design should be planned to provide radiation levels below the occupational limits for controlled areas and below the nonoccupational (for public) limits for noncontrolled areas.

The NRC requires that each cobalt facility possess or have available for use a survey meter capable of detecting exposure levels as low as 0.2 mR/h and as high as 1 R/h. The NRC further requires that the calibration of these instruments be checked at least once per year. We recommend that these same rules be followed for linear accelerator facilities.

D. Engineering aspects for patient protection

Other protection issues relate to the safe and reliable operation of the linear accelerator. In addition to radiation hazards that have recently been documented,³⁻⁸ mechanical hazards can also arise during normal operation of the machine. Ideally, an in-house engineering staff dedicated to the maintenance of the linear accelerator would ensure the safe operation of the unit. However, many hospitals may not be able to justify the expense of such highly specialized personnel on a full-time basis. In that case, a contract with the vendor or a

service organization to carry out routine preventive maintenance in addition to emergency repairs is necessary. This code of practice recommends that a vendor-outlined preventative maintenance program be carried out twice a year.

III. ACCEPTANCE TESTING

A. Overview

The acceptance of a radiotherapy accelerator requires a qualified radiation oncology physicist to determine that all applicable radiation safety standards are met or exceeded and that the machine meets or exceeds the contractual specifications. Published descriptions of the acceptance testing and quality assurance process^{1,20-29} must be extended to encompass the new technology of computer-controlled machines. Unlike their electromechanical predecessors, they are subject to change due to software changes also. In essence every software update produces a "new" machine. Each of these machines must be subjected to an appropriate subset of acceptance test procedures (ATP). The present document will therefore (1) summarize the testing requirements and the applicable methods and resources for the acceptance testing of traditional electromechanically controlled accelerators and (2) describe some extensions to the traditional procedures necessitated by computer-controlled accelerators.

The acceptance testing process, as it relates to the specifications, should be developed prior to purchase of the accelerator. Ideally, an institution should write a purchase order that would clearly state the accelerator's make, model, performance specifications (mechanical, electrical, radiation, safety interlocks, etc.) and a description of the means by which the machine will be tested for adherence to those specifications. This might be as straightforward as listing in the purchase order: (1) a manufacturer, (2) a model, (3) manufacturer's specification document (by part number), and (4) manufacturer's ATP manual (by part number) in the purchase order. However, in the event that the purchasing institution requires other than normal specifications for the machine (e.g., "tighter" mechanical specifications for a machine to be used for stereotactic radiosurgery), the third item above must be modified to reflect those tighter specifications. Similarly, if the purchasing institution wishes to replace, modify and/or extend the vendor's ATP (e.g., adding a series of interlock tests for a computer-controlled machine), then the fourth item above must be modified to reflect the intent. Care should be taken to assure that the parties are using a consistent set of definitions for technical terms.^{22,23}

Clearly, significant input from a qualified radiation oncology physicist is required prior to placing the order. In particular, the physicist should review and formulate not only the machine's specifications, but also an ATP for that machine prior to writing the purchase order. In this process, the manufacturer's experts should be considered as a resource. They should be asked to submit their ATP documents for evaluation. The physicist should decide whether the manufacturers ATP documents are sufficient or whether additions are necessary. Again, agreement to the specifics of the ATP prior to purchase is essential. Although numerical values for acceptance criteria are a matter of agreement between the vendor

and the purchaser, they must meet all nationally (and internationally) recognized minimum standards.²¹⁻²³

Acceptance testing serves three purposes: (1) it provides the mechanism by which the institution determines that it received what it intended to purchase; (2) it assures the safety of the patients and machine operators; and (3) it provides critical baseline data for future quality assurance reviews. The reader is referred to the forthcoming report of AAPM Task Group No. 35 (Ref. 29) for a detailed discussion of accelerator safety. Often it is necessary and/or useful to perform tests during acceptance that go beyond verifying the contractual specifications. It is important to structure the ATP so that any deficiencies that are discovered and rectified have minimal impact on tests previously accomplished. Consequently, once this structure has been defined it is essential that the tests be performed in the order specified.

It is important to note that it is the vendor's responsibility to prepare the machine fully for acceptance testing. After the safety tasks cited in Sec. III B (below) are accomplished by the physicist and installer, the installer should then complete the machine's installation. Definitive acceptance testing should be done after the installer takes the accelerator through a complete dry-run acceptance test on his/her own. This sequential system may prolong installation and acceptance testing time in some cases.

It is in the interest of both the manufacturer's representative(s) and the physicist(s) to work together to assure that the accelerator meets its specifications. Measuring equipment may be shared.

Many of the following subsections correspond to parts III, IV, and V of the previously published 1975 code of practice ('75 CP) for x-ray linear accelerators.¹ When the reader is referred there for additional details, the specific subsections will be designated as "('75 CP, subsection)."

B. Checking the treatment area

1. During installation

During the installation of the accelerator, the physicist shall assure that the facility is properly prepared. This task includes, but is not limited to: (1) installing proper radiation warning signs; (2) providing appropriate in-service training for those whose work involves the machine (e.g., pointing out the location of the emergency off switches); (3) assuring that appropriate warning lights are installed and are connected to the machine's console; (4) assuring that appropriate audio and video equipment is installed (to provide for monitoring of the room during installation); (5) assuring that appropriate door interlocks are in place and connected properly to the machine's interlock system(s); and (6) assuring that emergency power failure illumination is installed or available (e.g., flashlights).

2. At first delivery of beam

As soon as the accelerator is capable of producing radiation, several tests should be conducted to assure the safety of all concerned. These tests should include, but are not limited to: (1) validation of the proper operation of the door interlock system(s); (2) validation of the proper operation of the

emergency off switches; (3) a preliminary calibration of the machine output in all modes; (4) a determination of radiation exposure levels in areas occupied during these first beam-on measurements; (5) a determination of radiation levels outside the primary and secondary barriers under "worst case" conditions for the highest energy x-ray mode (at the highest available dose rate).

A full radiation safety survey will still have to be completed after a full calibration (see Sec. II above). The preliminary survey is necessary to assure that as installation and acceptance continue, the environment is safe for occupancy.

C. Initial checking of mechanical and radiation systems

The following tests are best performed in the order presented here, as the validity of each subsequent test depends upon the accelerator having passed the preceding test.

1. Alignment of collimator axis and collimator jaws

The adjustable collimator jaws at closure must maintain symmetry about the collimator assembly (mechanical) axis during rotation of the collimator assembly. This can be tested with a mechanical front-pointer grasped by all four jaws and extending the pointer toward the isocenter. Rotation of the collimators will enable the end of the front-pointer to trace out (on a sheet of paper at the isocenter) any misalignment between the collimator axis and the closing faces of the jaws. ('75 CP, IV.A.)

If a front pointer is not provided by the manufacturer, one can be easily fabricated and used. However, if clamping a front pointer between the jaws is not advisable (for example, in the case of independently driven jaws) or not possible (for example, if there is a protective plastic cover preventing access) then the checking of the collimator jaw symmetry can be performed indirectly by ensuring symmetry of the light field as described in the next section.

2. Collimator axis, light localizer axis, and cross hairs

The axes of the collimating system and the light field must be congruent, within the specified tolerance over all ranges of motion, for all mechanical systems. The image of the cross hairs must project along this common axis. The light field center is compared to the mechanical center determined above, the light field edges are determined to be symmetric about the center (and adjacent edges are determined to be perpendicular to one another), and the cross hair location is determined to lie on the mechanical axis. If any adjustment is needed generally the light source position is adjusted to obtain congruence of the light-field axis with the mechanical axis before the cross hairs receive their final adjustment. ('75 CP, IV.B.)

3. Light field and radiation field congruence and coincidence

a. Light field and radiation field symmetry. Having verified that the light field axis and mechanical axis are congruent, it is now necessary to verify that collimator jaws are symmetric by exposing two films with collimators rotated

180° between exposures. Next the congruence of the radiation field axis with the light field axis is verified. This congruence and symmetry must be verified over the full range of both collimator and gantry positions. These tests are generally accomplished with a series of x-ray films (often “ready-packed”) placed at the isocenter with the film plane perpendicular to the light field axis. Marks placed on the films denoting the light field center and edges are compared, after exposure and development, with the radiation field center and edges. The beam edge is usually defined as the line of 50%-of-central axis dose which is practically the same as the line of 50%-of-central axis optical density of film used. The full width at half maximum (FWHM) of the radiation field is generally measured using optical density (OD) which therefore requires that the film’s OD be within the linear range (or that the film’s dose response be independently determined). (‘75 CP, IV.C.1.)

b. Light field, radiation field and field readout agreement, and accuracy. It must be demonstrated that the light field size and readouts for the collimator field size are in agreement with the size of the radiation field over the full range of collimator and gantry rotations. This is generally accomplished with film or with scanning dosimetry systems using a water phantom. Determinations should be made over the full range of field sizes and at different SSDs. The distances from the center to edges should be equal for the light and radiation fields and should agree with the values displayed on the readout system(s). (‘75 CP, IV.C.2.) It should be verified that the mechanical and digital readouts agree with each other.

4. Mechanical isocenter location

The location and size of the volume containing the isocenter (the idealized intersection of the collimator, gantry, and couch rotation axes) must be determined for the full range of collimator, gantry, and couch rotations. Typically, the specifications for the purchase of the accelerator will state an upper limit for the radius of a sphere that contains the “intersection” point for all machine orientations. For clinical use, the isocenter is determined using the optical system, which is accomplished by projecting the optical cross hair image onto and beyond a variety of mechanical jigs placed at the supposed isocenter. (‘75 CP, IV.D.)

If the manufacturer provides a “front pointer,” it should be confirmed that it accurately points to the isocenter location. Any additional distance marks on the front pointer can be verified at this time.

The accuracy of the optical distance indicator at the isocentric distance must be verified. Verification of other distances can be accomplished during the couch readout tests.

If the machine has a beam stopper, the final stated volume (3 axes) of the mechanical isocenter should be determined with the beam stopper in its extreme positions (in/out).

5. Radiation isocenter location

Once the size of the bounding sphere for the mechanical isocenter has been shown to be within specification, the location and size of the radiation isocenter bounding sphere

should be determined. These tests are generally executed semi-independently for components of the accelerator that can rotate.

a. With respect to the collimator axis. In a plane perpendicular to the central axis, the radiation beam center should remain within a circle of some specified diameter at isocenter for the full range of collimator rotation. This circle should also include the mechanical isocenter. This is generally demonstrated with “star-shot” films, one film for the upper and one film for the lower jaw. (‘75 CP, IV.E.1.)

b. With respect to the treatment table. In a plane perpendicular to the central axis, the radiation beam center should remain within a circle of some specified diameter at the isocenter and within a plane for the full range of rotation of couch around the isocenter. This circle should also include the mechanical isocenter. This is generally demonstrated with a star-shot film on which the location of mechanical isocenter can be pin-pricked utilizing the front pointer. (‘75 CP, IV.E.2.)

c. with respect to the gantry. The intersection of the central rays of a series of beams, each directed from a different gantry angle, should lie within a circle of specified diameter in a plane containing these central rays. This circle should also include the mechanical isocenter. This is also generally demonstrated with a star shot film. (‘75 CP, IV.E.3.) One should also check the isocentricity of the gantry in the direction parallel to the gantry axis of rotation.”

D. Other mechanical system tests

1. Patient support system

The couch must be tested to assure that: (1) its flex in both longitudinal and lateral travel of the couch with and without load is within specification and that the couch can lift the specified maximum load; (2) the couch’s brakes are able to lock in any position; (3) the vertical speed of the couch under load is within specification; and (4) couch position readouts are within stated accuracy.

This is often a convenient opportunity for verifying distances other than source-to-axis distance (SAD) on the front pointer(s), and for verifying accuracy of the optical distance indicator system. That is, their accuracy can be determined at the time the couch vertical motion measurements are being made.

2. Anticollision system(s)

Because of the wide variety of techniques used for anticollision devices, it is not possible to recommend specific procedures. Depending on the quality and detail of the manufacturer’s ATP document and the specifications agreed upon in the purchase order, the physicist may have to devise appropriate tests. The anticollision system(s) should be tested over the full range of motion of the collimators, gantry, and couch.

3. Beam modifier systems (electromechanical aspects)

Because of the wide variety of mechanisms used for beam modification, it is not possible to recommend specific proce-

dures for all possible devices. Depending on the quality and depth of the manufacturer's ATP document (and the specifics of the purchase order) the physicist should devise the tests used to verify that the mechanisms perform as set forth in the manufacturer's specifications and the purchase order. The physicist should verify that the devices mount as specified, move as specified (range and speed), and that any associated readouts are linear and accurate.

a. Electron applicators (cones). Generally, each electron beam cone requires a particular setting of the accelerator's adjustable photon collimators, which may vary with electron energies. The mechanisms for assuring the proper collimator setting vary. It is, however, necessary to verify that electron beams cannot be produced without the correct collimator settings (within agreed upon tolerances) for each cone size/electron energy combination.

4. Beam stopper (electromechanical aspects)

If the accelerator is equipped with a retractable beam stop its drive mechanism should function over the full range of gantry positions. The beam stop must meet its travel time specifications. If a "pendulum" or other electromechanical system is used to prevent beam-on for a range of gantry positions, the physicist must demonstrate that those mechanisms are functioning as specified. The maximum field size interlock with beam stopper must be verified. Computer-controlled interlocks will be discussed in Sec. III E2.

E. Console system tests

A radiotherapy accelerator may be controlled electromechanically, or by computer, or by a combination of both methods. Nevertheless, there are items that can be tested in a similar way regardless of the type of control mechanisms. In addition to the main console there may be auxiliary displays and/or control stations. Acceptance tests should include these peripheral devices. A "record and verify" (R&V) system may be part of, or attached to, a console. The physicist must verify the proper operation of that system and particularly the interaction of the R&V system with the console (especially if the R&V system can be used to program the accelerator).

Computer-controlled accelerators pose special problems for the physicist. Not only are several additional tests required during acceptance testing,^{26,29} but because software can be changed or upgraded, appropriate acceptance testing should be repeated when the manufacturer provides hardware and/or software updates.

1. Mode selection

Operation of a multimodality accelerator must not permit dangerous combinations of beam current, target, and filter for photon and electron beams.⁸ The physicist must understand the methods by which this is accomplished for his/her accelerator. Tests must be devised and run to demonstrate that the control and safety systems are functioning properly. Such tests should include simulations of possible "mistakes" of any known kind.⁷ For a computer-controlled accelerator, these tests should be repeated whenever a software update is installed.

2. Computer-controlled accelerator software validation

The accuracy of software for a computer-controlled accelerator must be validated during acceptance testing. Any changes to that software (for example, a manufacturer's update) can cause the accelerator to function differently than as expected. The physicist must assure that the software currently provided by the manufacturer is the one that is in use. Software updates must be subjected to an ATP.

During the software ATP, the physicist must assure that the controlling software causes the accelerator to meet: (1) its operational specifications; (2) its safety specifications; and (3) its dosimetric specifications.

Lastly, the physicist should determine that there is a mechanism for assuring day-by-day validity of the software (e.g., by observing that the manufacturer's checksum routines function properly when the executable code is loaded into RAM). (A checksum, in its simplest form, is a number calculated by adding the contents of every byte of instruction and data in a "program" regardless of how the contents of that byte were intended to be used. It serves to demonstrate constancy, i.e., that the program and the data have not changed.)

a. Operational specifications. The manufacturer should provide the physicist with a functional specification for the accelerator's operation. It then becomes the physicist's responsibility to verify that the controlling software causes the accelerator to operate according to the specification. For example, the specifications might state that a password is required before the accelerator can be placed in a high-dose-rate mode. The acceptance testing process should verify that this is the case.

b. Safety specifications. Any safety related systems must be tested over a reasonable range of test parameters. For example, if the software determines collision possibilities according to couch and gantry positions, then the manufacturer should inform the physicist as to the function and structure of the algorithm so that it can be verified during acceptance testing. If, for example, the software monitors dose per pulse, then the manufacturer should provide some means for varying the input to the control system (i.e., simulating an excess dose per pulse) so that the software's proper response to an out-of-bounds value can be verified.

The manufacturer should provide information regarding the checking of computer-controlled interlock systems, as described in Sec. III G.

c. Dosimetric specifications. The proper operation of the monitor unit integrators, the symmetry meters, the dose rate circuits, etc. must be verified regardless whether or not they are software controlled. If computer control for dosimetric performance is involved, the manufacturer should provide the physicist with details as to the algorithms or functional specifications so that the system's response can be verified.

If the accelerator is calibrated through software, i.e., the number of cGy per monitor unit is adjusted by entering numbers at a keyboard, it should be assured that these calibration factors are correct, properly "loaded" each time the accelerator is used, and secured from unintentional changes.

d. Validation of software. Manufacturers should provide

a mechanism for validating system software prior to accelerator operation (for example, the system could calculate a checksum of the executable image of the software and compare the resulting value to an independently stored "correct" value). The manufacturer should provide this safeguard so that the physicist can observe and confirm its correct function.

e. Backups. Computer-controlled accelerators should include some system for backing up programs and data. A physicist should verify that these backup systems work properly. This can be accomplished by doing a full (and partial and/or incremental) backup followed by a full and partial restoring of the software. Presumably, the manufacturer's software will include validation mechanisms (such as checksums on files) that can verify that the restored files and software have not been corrupted by unintentional changes.

3. Readouts

All readouts (monitor units, collimator field size, optical range finder, couch position, etc.) must meet their specifications for accuracy and linearity. Some of these tests can be accomplished during the initial tests cited above. For example, during the light field/radiation field coincidence determinations (Sec. III C3), the collimator field size readouts can be recorded for comparison with the measured light and radiation field sizes. This and other readout tests should be accomplished over the entire range of motion of the subsystem in question. Care should be taken to demonstrate that any auxiliary readouts authentically duplicate the primary readout.

4. Record and verify systems

There are presently two broad classes of R&V systems: (1) Those that simply maintain a record of the patient's treatment parameters; and (2) those that program the accelerator as well as maintain a record of treatment. The physicist must determine that the recorded treatment information agrees with the actual values for all modes of operation. If the R&V system is capable of programming the accelerator, the physicist must determine that the accelerator is properly programmed for treatments within the range of possible combinations of treatment parameters. Specific guidelines about the necessary tests for these purposes are yet to be developed by the AAPM.

F. Checking of radiation systems and beam parameters

1. Beam output

a. Preliminary calibration. A preliminary calibration of the beam output shall be performed using the methodology of the "TG-21" protocol³¹⁻³⁴ or IAEA protocol.^{34,35} It is often advisable to perform the accelerator calibration early in the acceptance testing process, so that long-term stability of the calibration may be tested. Final beam calibration should be performed and verified independently before commissioning the machine as described in Sec. IV below.

b. Adjustability and range. The mechanism (trim-pot, software, or "other") for adjusting the number of cGy deliv-

ered per monitor unit (MU) should have sufficient range. This can be assured by varying the cGy/MU adjustment and measuring the accelerator's output.

c. Stability. The cGy/MU adjustment should be stable over both the long and short term. Long-term stability can be demonstrated by performing an output calibration as early as possible in the acceptance test process, and by repeating that calibration daily during the acceptance testing period. Short-term testing (over time scales of minutes, hours, or days) can be accomplished at any time during acceptance testing. Stability testing should be done for all operational modes.

d. Timer. The timer associated with the integrator systems must be accurate, linear, and capable of turning off the radiation beam when the programmed amount of time has passed. Generally, this can be tested by running the the accelerator for a range of times, with the programming set to cause termination of beam on time rather than dose, while comparing the duration of the run with the time recorded by a stopwatch. ('75 CP, IV.F.)

2. Monitor characteristics

a. Linearity and end effect. The behavior of the monitor system with respect to integrated dose should be checked. This can be accomplished by measuring the accelerator's output for programmed doses covering the full range of MU settings. That is, the accelerator's output should be determined for MU settings of, e.g., 10, 20, 50, 100 200,..., and the result plotted versus MU set. In some accelerators MU settings below 10 go through port filming circuits rather than treatment circuits. For such machines, linearity check for less than 10 MU can lead to problems and only MU greater than 10 should be used, as it is appropriate for the treatment. A linear regression analysis will show any significant deviation from linearity, and the intercept on the abscissa from this regression will provide the monitor system's end effect.³⁶ Alternatively, the end effect can be determined by the "two exposure method."³⁷ When comparing linear regression and two exposure method it should be noted that a negative intercept is the same as a positive alpha, i.e., negative intercept of 0.1 MU means the same as a positive alpha of 0.1 MU. To perform the linearity check one must ensure that the ion chamber system used is linear. It should be noted that the end error correction is generally small for modern accelerators and may be positive or negative.³⁸

b. Dose rate accuracy. The console/monitor system should assure that the actual dose rate (in MU/unit-time) is within the specified tolerance of the programmed value. This can generally be determined with an independent timer.

c. Dose rate dependence. Some single, low energy machines operate on a single dose rate. However, many computer-controlled multimodality machines offer several dose rates. For such machines the behavior of the monitor system with respect to dose rate should be determined. This can be accomplished by measuring the accelerator's output for programmed dose rates covering the full range of MU/unit-time settings.

d. Constancy of output with gantry position. It should be demonstrated that the accelerator's output is not significantly

affected by gantry position. This can be accomplished by a series of in-air measurements (with appropriate buildup cap) of the machine's output for various gantry angles.

e. Monitor chamber seal integrity. If the accelerator is equipped with a sealed monitor chamber system it is necessary to monitor the accelerator's output(s) versus ambient pressure and temperature. Any indication that the accelerator's output(s) are tracking the room pressure and temperature variations is cause for questioning the integrity of the monitor chamber's seal.

f. Pressure and temperature compensation. If the accelerator is equipped with an unsealed monitor chamber that has an automatic pressure and temperature compensation system, it must be demonstrated that the compensation is correct and stable. This will again involve tracking the accelerator's output versus ambient pressure and temperature.

g. Collection efficiency. The monitor chamber's collection efficiency should be high enough so that no significant deviation from the accelerator's calibration occurs at even the highest dose rates used. If control of the high voltage applied to the monitor chamber is accessible, the collection efficiency can be directly determined, if so desired.³⁹ The latter, however, is not necessary if dose per monitor unit has been verified for all available dose rates on the machine.

If the accelerator compensates, perhaps in software, for a known ion collection efficiency at high dose rates, the physicist should verify that the algorithm and its implementation is correct (see Dose rate dependence, Sec. III F 2 c above).

h. Backup counter. Proper operation of the MU counter designed to preserve a record of monitor units delivered prior to a power failure should be verified.

3. Flatness

Flatness can be specified as a maximum permissible percentage variation from the average dose across the central 80% of the full width at half maximum (FWHM) of the profile in a plane transverse to the beam axis. That is, the flatness F is given by

$$F = \frac{M - m}{M + m} \times 100\%, \quad (1)$$

where M and m are the maximum and minimum dose values in the central 80% of the profile.

Flatness is usually specified for one or more field sizes, for a particular depth in a phantom, and for several gantry angles. Some manufacturers also specify flatness along the diagonals of the beam(s). In addition to specifying flatness at a depth (e.g., 10 cm), it is also advisable to have specification at d_{max} .

Generally, flatness is measured in a water phantom with a dosimetry scanning system. Alternatively, film may be used. ('75 CP, IV.G.) While manufacturers generally specify flatness for the two transverse directions along each of the collimator motions, it is recommended that the physicist check flatness for major planes (i.e., in-plane, cross-plane, diagonal-plane, etc.) that contain the collimator axis. This is most easily accomplished by (film or water phantom) scanning in a plane perpendicular to the collimator axis (at depths of d_{max} and 10 cm) with the subsequent display of isodose

lines. This "contour map" of dose in the perpendicular plane will alert the physicist to any flatness deviations.,

4. X-ray off-axis ratios (horns)

Some manufacturers specify the maximum dose along a cross beam profile relative to dose on the central axis for the maximum field size at the depth of maximum dose on the central axis d_{max} . The confirming measurement is most easily accomplished with a water phantom scanning system. While manufacturers generally specify horns for the two transverse directions along each of the collimator motions it is recommended that the physicist check in all major planes that contain the collimator axis. This is most easily accomplished by (film or water phantom) scanning in a plane perpendicular to the collimator axis (at depths of d_{max}) with the subsequent display of isodose lines. This contour map of dose in the perpendicular plane will alert the physicist to any excessive horns. It should be noted that IEC has suggested limits on off-axis ratios (horns).^{22,23} Also, the NACP protocol recommends that the uniformity index (defined as the ratio of the area enclosed by the 90% contour to that by 50% contour in a reference plane) be greater than 0.80.

5. Symmetry

The data obtained above can be used for the determination of beam symmetry. The definition of beam symmetry should be agreed upon prior to purchase as it varies from manufacturer to manufacturer. Symmetry is often defined as a maximum permissible percentage deviation of the "left-side" dose from the "right-side" dose of a beam profile often at 80% of the FWHM points.

Symmetry is usually specified for both transverse directions, each along the direction of collimator motion, for several field sizes, for a particular depth in a phantom, and for several gantry angles. Some manufacturers will also specify symmetry along the diagonals of the beam(s).

Generally, symmetry is measured using a water phantom scanning system. Many of the current computer-controlled scanning systems provide software to determine symmetry using a wide variety of methods. The accuracy of such software should be checked. Alternatively, film may be used for these measurements. ('75 CP, IV.G.) If the accelerator's symmetry meter is demonstrated to be accurate, it can easily be used to measure symmetry at various gantry angles. While manufacturers generally specify symmetry in two directions only, it is recommended that the physicist check symmetry for all major planes that contain the collimator axis. This is most easily accomplished by (film or water phantom) scanning in a plane perpendicular to the collimator axis (at depths of d_{max} and 10 cm) with the subsequent display of isodose lines. This contour map of dose in the perpendicular plane will alert the physicist to any excessive asymmetries.

6. Penumbra

The penumbra, generally defined as the lateral distance between the 80% and 20% of maximum dose points on one side of a beam profile, must be within specification (if any). The data for this determination are usually part of the data

acquired for the flatness and symmetry checks. As is the case for those checks, measurements should be made in both transverse directions and for a variety of field sizes. If ionization chambers are used for determination of penumbra, allowance should be made for the finite size of the probes.⁴⁰⁻⁴² Film is a better detector for this task. The film scanning system should not spread out a block edge by more than a quarter of the minimum expected penumbra.

7. Energy of x-ray beams (d_{\max} and % depth dose)

X-ray beam energy can be specified in terms of depth of d_{\max} and percent depth dose at another depth for a 10X10 cm field. For example, a nominal 18 MV beam may be specified as having d_{\max} at 3.3 ± 0.2 cm and a % depth dose of $80\% \pm 1.0\%$ at 10.0 cm depth in water. The physicist must assure that all x-ray beams have the appropriate percent depth dose for those field sizes for which specifications exist. Usually the manufacturer's specifications of the nominal energy are not consistent and cannot be verified easily. These measurements are most easily carried using a water phantom scanning system.

8. Collimator transmission

The physicist must verify that transmission of the x-ray beam through the adjustable collimators is no greater than the specified value. This measurement should be made independently for the upper and lower jaws. The measurement can be made by reference to the maximum in-phantom dose on the central axis of the largest field, and by placing an ionization chamber with appropriate buildup in the shadow of one set of jaws, with one set of jaws open and one set closed. Stem (and cable) effects should be accounted for when taking the reference measurement at large field sizes.

9. Energy of electron beams (% depth ionization)

Electron beam energy can be specified in terms of the depth of 80% and 50% depth ionization (both distal to the depth of maximum dose, d_{\max}) for a 10X10 cm field. For example, a nominal 12 MeV electron beam may be specified as having a depth of 80% of max at 3.8 ± 0.1 cm and a depth of 50% of max at a depth of 5.2 cm. The physicist must verify that all electron beams meet the specifications. These measurements are most easily carried out in a water phantom scanning system. If a cylindrical ionization chamber is used, then a chamber displacement correction must be made to the point of measurement.

10. X-ray contamination

The data acquired in Sec. III F9 above can be used for this determination if the percent depth ionization curves are carried to sufficient depth. Then, it must be assured that the x-ray contamination of each electron beam is within the specified limits. For example, this specification may take the form that, in a water phantom at a depth of 10 cm beyond the practical range, the ionization shall be less than some specified percent of the maximum. It should be noted that there are differing definitions of x-ray contamination be-

tween vendors. The exact definition and measurement method needs to be agreed upon prior to purchase.

11. Rotation and arc therapy using photons

a. Dose per unit angle. It must be demonstrated that the programmed dose per unit angle is independent of the size of arc and the particular start and end points of arc and the total dose. These measurements can generally be accomplished by mounting a chamber with appropriate buildup cap at the isocenter coaxially with the axis of rotation and performing a series of arc irradiations around the chamber.

b. Arc termination. It must be demonstrated that the accelerator stops at the proper angle and that the cessation is due to either delivery of the programmed total dose or arrival at the stop angle. The physicist should verify that both (or any other manufacturer supplied) mechanisms for arc termination are operating as specified. Additionally, if the accelerator stops "on-angle," the delivered dose should be within the specified bounds of the programmed dose. Conversely, if the accelerator stops "on-dose," the extent of rotation should be within the specified bounds of the programmed arc. These determinations can generally be made during the dose measurements described above by monitoring the dose integrator and stop-angle readouts at the console, providing that their accuracy, repeatability, etc. have already been verified.

12. Beam modifying devices

a. Wedges. Usually there are wedges for angles of nominally 15° , 30° , 45° , and 60° . For each wedge angle, there are usually two identical wedges for opposite directions (toe-in and toe-out; toe-left and toe-right). Mechanical play in the slide direction is bound to develop over time. Its influence on the wedge factor can be eliminated by mounting wedges perpendicular to the slide direction. It is necessary to demonstrate that the wedge attenuation factor (see Sec. IV C2 for details) is correct for any insertion direction (or for any wedge if different physical wedges are used for the different insertion direction). Ideally, this would be assured by the use of a point-detector located on the beam's central axis. Practically, cylindrical detectors are often used and must be oriented so that their longitudinal axis is perpendicular to the slope of the wedge and parallel to the mounting tray. Furthermore, the detector's sensitive volume must be placed on the beam's central axis. That the radiation detector is on the collimator rotation axis can be checked by rotating the collimator 180° with a 45° wedge in place. If the reading varies by more than 1%, the chamber should be moved until this tolerance level is observed. This test should be done at several gantry orientations. A change in the wedge attenuation factor could indicate lack of consistency in the wedge-tray positioning in the accessory mount. Wedge transmission factors need to be measured for several field sizes and depths.

Wedges must also be tested to assure that whatever mechanism is used to prevent jaw settings larger than the wedge size functions properly. The nature of this test is dependent on the particular mechanism in use and to some extent on whether the wedge is mounted on an opaque or transparent carrier.

b. Asymmetric jaws. It must be demonstrated that the jaws move properly; that the readouts are accurate; and that congruence of light field and radiation field is maintained through the full range of motion of the jaws (including any axis crossing). Tests to demonstrate these qualities should be run for a range of gantry and collimator positions. Otherwise, the tests are similar to those already described.

c. Moving jaw wedges. For those accelerators capable of simulating the presence of a wedge with a programmed movement of a collimator jaw, the physicist must verify that the resultant dose distributions are reproducible and reasonable facsimiles of those that would be expected of a "real" wedge. Tests should include dose distribution measurements for a range of gantry and collimator positions. Furthermore, the long-term reproducibility of the dose distributions produced by the moving jaw wedge system should be demonstrated.

d. Beam stopper. It must be verified that the transmission through the beam stopper is as specified. It must further be verified that the accelerator will not run unless the beam stop is extended for specified gantry angle zones. These tests are not necessary if the shielding walls have adequate thickness to reduce radiation levels outside the room to an acceptable value.

13. Isodose (isoionization) curves

Sample isodose curves for x rays (including some with wedges) and isoionization curves for electrons should be obtained to demonstrate that the accelerator is meeting its isodose (isoionization) specifications. Sample curves should be available from the manufacturer. These measurements are generally made with a water phantom scanning system. Film dosimetry is a viable option for these measurements.

14. Surface Dose

It must be assured that the "surface" (or "skin" or "entrance") dose value(s) meet the specifications. This determination assures that beam contamination is within acceptable limits. These measurements are usually accomplished with plane-parallel ionization chambers or TLDs. For surface dose measurements, the plane parallel chambers need a correction factor that can be determined using an extrapolation chamber.⁴³ Extrapolation chambers are particularly well suited to this task and are the gold-standard for this purpose. The differences in skin dose from solid versus pattern-drilled versus "hollow" block trays should be evaluated before use.

G. Checking interlock systems

The accelerator's interlock system(s) must be thoroughly checked. The acceptance testing procedure should contain the method by which each interlock shall be tested. Attention should be paid to the hierarchy of control as specified by the manufacturer's operating manual. In older accelerators the interlock circuitry is often composed of microswitches and relays. It is relatively easy to understand and test these circuits. In somewhat more modern accelerators, digital circuits (but not necessarily computers) are used in the interlock systems. For example, on some accelerators, binary coded deci-

mal (BCD) parallel signals from plugs on wedges have to match those from console switches during programming and operation. Again, it is relatively easy to understand and test these circuits. In many modern accelerators, computers (and hence software algorithms) interact with or control the interlock system(s). Generally, the software is proprietary and not available for inspection. This is not unreasonable as it preserves trade secrets, and in any case the source code listings would be of little use to the majority of physicists (it would take considerable effort by a specialist to fully understand the software control logic). It is therefore very difficult to understand the internal structure of, and to test, these hardware/software interlock systems. Nevertheless, it is necessary in order to assure patient and operator safety.^{26,29}

A testing method that works for computer-controlled accelerators has been described^{26,29} and is summarized here. In order to test the interlocks, the physicist will require the interlock system(s) functional specifications from the accelerator's manufacturer. These should clearly detail each interlock in terms of the fault detection sensor, method of comparison (hardware, software, or both), trip conditions, and desired result. For example, a monitor chamber power supply interlock might be described as voltage detection at circuit board 7 pin 14; comparison of digitized voltage to range of allowed values by software; interlock tripped if the absolute value of voltage is less than 500 V, if interlock is tripped the accelerator shall not be capable of producing beam and if the beam is already on, the accelerator shall shut down. With this type of functional specification in hand, it becomes possible for the physicist and the manufacturer's representative to test the systems without having to know additional (internal) details. For the example cited above, they could test the interlock by "reducing" the voltage at the sensing point. It should be emphasized that simulation of a fault should be accomplished as close as possible to the sensor, and that interlock system(s) testing should be done in the same operational mode(s) as are used to treat patients.

Clearly, it would be advantageous to all if the manufacturers provided instructions and materials for function testing of interlock system(s) in their ATP. Physicists can assure the availability of these resources by including them as a requirement in the purchase order. The continuing need for these resources is apparent when one considers that any software-controlled machine should be, to some degree, re-accepted (including all interlock testing) whenever a software update is installed.

H. Multileaf collimators

As multileaf collimators differ in design from accelerator to accelerator, a detailed description of their acceptance testing is necessarily beyond the scope of this document. Nevertheless the physicist must assure that a multileaf collimating system meets specifications and all applicable regulatory requirements. Items that must be tested include but are not limited to: (1) actual leaf positions versus programmed leaf positions for each leaf; (2) time to reach position (a critical parameter for dynamic conformal therapy); (3) radiation leakage through the leaves; (4) radiation leakage between the leaves; and (5) beam contamination (caused by scattering from

the leafs). Many of these tests must be repeated at several gantry angles to assure that the changes in mechanical stresses at different angles do not cause the system to function outside of its specifications.

I. Checking ancillary equipment

Any significant ancillary equipment or software purchased with an accelerator should be acceptance tested to assure that it meets specifications. Such items include patient positioning lasers, electron beam mold kits, high dose rate therapy software (as an option to the computer control system), etc. Test procedures for each of these items, and any other items offered by the manufacturer, should be part of the ATP document provided by the manufacturer. If not, the physicist should devise suitable tests.

J. Summary of acceptance testing

Defining the ATP in detail should be considered a part of the accelerator selection and purchase process. The manufacturer should be asked to submit its ATP documents along with its accelerator-specification data sheets for consideration. These documents should be reviewed for completeness, agreement on the definition of terms by the IEC^{22,23} and suitability to the institution's needs. If the manufacturer's ATP documents (and materials) are incomplete, the purchasing institution should add whatever is felt to be necessary to the purchase order. These efforts, undertaken early in the purchase negotiations, can save considerable time and effort in acceptance testing.

IV. COMMISSIONING

A. Overview of commissioning a radiotherapy accelerator

A satisfactorily completed acceptance test simply assures that the accelerator and its associated systems satisfy all performance specifications and pertinent safety requirements. Good radiotherapy also requires the ability to determine accurately the absorbed dose at any point of interest in the patient. Therefore, it is necessary to have the ability to perform treatment planning, which is the process of selecting the appropriate radiation modality and treatment technique for individual patients. Specifically, it refers to the geometric and physical parameters needed to deliver the prescribed dose and dose distribution to the target volume. After successful acceptance testing, additional data are needed to enable proper planning and delivery of radiation treatment using an accelerator. A comprehensive set of beam data must be acquired and entered into the radiotherapy treatment planning (RTP) system. "Commissioning" refers to the process whereby the needed machine-specific beam data are acquired and operational procedures are defined. It includes but is not limited to: (1) beam data acquisition; (2) entry of beam data into an RTP system and testing of its accuracy; (3) development of operational procedures; and (4) training of all concerned with the operation of the new accelerator.

Generally, there is a great pressure to initiate clinical treatments as soon as the acceptance testing has been com-

pleted. Rushing into clinical implementation without completing the tasks described in this section can potentially cause harm to the patients. Therefore, an appropriate time must be scheduled for the proper commissioning of a radiotherapy accelerator. The length of time needed depends on many factors, such as availability and experience of personnel and proper instrumentation and type of accelerator. If proper instrumentation, in particular a scanning water phantom system, is available, a single energy photon machine can be commissioned in about 2-4 weeks, depending on equipment and personnel. The physicist with less sophisticated equipment may need more time for commissioning. On the other hand, a multimodality accelerator with two photon energies and several electron energies can take about 6-8 weeks of intensive effort (requiring 16-h shifts) to commission if the clinical needs require a faster schedule than more personnel should be hired and/or overtime work shifts should be planned. Partial commissioning, say of one photon energy, is another option that is not optimum and is therefore discouraged. It may also be possible to commission the accelerator for simple procedure first, such as the use of photons only. We strongly recommend that the commissioning process should be carefully planned and executed because it is vitally important for the safe and efficacious implementation of radiation treatment. It involves not only beam data collection but developing the means for proper planning and delivery of treatments using the new accelerator. The clinical implementation of a radiotherapy accelerator mandates the use of a dedicated computerized treatment planning system. Modern medical imaging techniques such as CT, MRI, and digitized x-ray films allow highly detailed target localization in three dimensions. State-of-the-art radiation treatment planning systems, therefore, should account for the three-dimensional aspect as well. It should enable interactive visualization of treatment techniques and relative distributions of dose associated with a proposed beam configuration and anatomy. Detailed recommendations in regard to the selection of a treatment planning system are beyond the scope of the present report. It is necessary, however, to be aware that any computerized treatment planning system requires a considerable amount of input data characterizing the radiation beams, the patient, and the treatment technique. The nature and volume of these data depend on the underlying dose calculation models and associated algorithms. Each modality and beam energy requires its own set of characteristic data. In general, the sets of characteristic data necessary for commissioning these systems are different from one treatment planning system to another. It is therefore mandatory when commissioning a machine for clinical use to acquire all the data needed by the RTP system to be used and to test the accuracy of this data set.

For the purposes of commissioning, one is mostly interested in relative measurements: depth ionization or dose curves, cross beam profiles, and isodose plots. Dosimetry measurements for acquiring beam data are best performed in water using appropriate radiation detectors. The essential features required of any measuring device are:

- (1) sufficient sensitivity;

- (2) stability;
- (3) negligible leakage;
- (4) energy independence;
- (5) sufficient spatial resolution, and
- (6) linearity.

The most convenient equipment for this type of measurement is an automatic radiation field scanning system. Typically, it consists of a large water tank and an automated field scanning support mechanism that can transport a measuring probe with great accuracy and precision. Generally automatic field scanning and isodose tracking are possible in the three principal planes. The measurements can be performed with a pair of small ion chambers or diodes, the signal from one of the pair, mounted at a fixed position in the beam, serving as a reference. Using the automatic radiation field scanning dosimetry systems in scanning electron (or photon) beams requires the use of signal integrating methods with integration times that are long relative to the field sweeping time. Alternatively, film dosimetry can be used in the case of scanning electron/photon beams.

Before clinical implementation of radiation treatments using an accelerator, it is essential to develop a method(s) for the determination of monitor units necessary to deliver a given dose at a reference point in the patient. It should be possible to manually calculate the monitor units necessary for simple treatments as well as to calculate monitor units necessary to execute a particular treatment plan developed using a computerized RTP system. At all times it must be possible to calculate, manually, the dose at some critical points such as a reference point at or near the center of the target, at points at or near the edge of the target, and at points in critical normal tissues. There are several approaches to the determination of monitor units necessary to deliver a given dose at a reference point in the patient. For a description of some of them, see the methods described by Johns and Cunningham,⁴⁴ Khan,⁴⁵ Van de Geijn,^{46,47} and Purdy.⁴⁸ The AAPM Radiation Therapy Committee has formed a new task group (No. 49) to develop guidelines for the determination of monitor units to deliver a given dose using one or more beams of photons or electrons comprising a specific radiation treatment.⁴⁹

The physicist should also figure out in advance of beam data acquisition how the institution is going to use single/double asymmetric jaws as well as multileaf collimators. Specific AAPM guidelines on asymmetric jaws are being developed by AAPM Task Group 52.

Finally, as a part of commissioning, a series of check measurements needs to be done to verify all data used in the calculation of dose for both electron and photon fields. The measurement of the dose in clinical field arrangements that includes all devices should be done to verify the treatment planning computer calculations and hand calculations.

B. Dosimetry calibration

The accelerator should be calibrated according to the methodology of the AAPM 1983 protocol³¹⁻³⁴ (also known as the TG-21 protocol) or the IAEA 1987 protocol.^{34,35} Calibration refers to the relationship between dose (cGy) and

monitor units (MU) at one particular depth, for a specific field size at a given value of source-to-surface and source-to-chamber distances. It is also advisable to obtain an independent check on the calibration of beam output from a new accelerator. This can be accomplished by using the mailed TLD service for both photons and electrons from the Radiation Dosimetry Services at the University of Texas Medical Center or the mailed Fricke dosimetry service for electrons from NIST.⁵⁰

It should also be noted that the 1983 AAPM protocol for dosimetry calibration of high energy photon and electron beams has been extended by AAPM Task Group No. 39 to address the special problems arising from the use of plane-parallel ionization chambers.^{51,52} Also, the 1983 AAPM protocol in its entirety is being reviewed by a new AAPM task group,⁵³ No. 51, which is likely to present its recommendations in about 4 years from the date of publication of this document.

C. Commissioning photon beams

One of the most important tasks in commissioning photon beams is selecting a method for dose calculations and then collecting the necessary beam data. Conventional manual treatment planning makes use of two basic ingredients: central ray depth dose data for square or rectangular beams and isodose charts. The central axis data may be in the form of percentage depth doses (PDD) for standard SSDs or tissue air ratios (TAR), or tissue maximum ratios (TMR) or tissue phantom ratios (TPR). While PDD data are applicable to the SSD used in measurements, the TAR, TMR, and TPR data can be used for variable SSD techniques. For a recent review of clinical photon beam dosimetry, readers are referred to a review by J. Purdy at the 1990 AAPM Summer School.⁴⁸ Also, the report of the newly formed AAPM Task Group 46 to tabulate accelerator-produced photon beam data would serve as a useful guide for checking the beam data.⁵⁴

The accuracy of computerized radiation treatment planning (RTP) systems for photon beams can be verified using the data provided by AAPM Task Group 23, which has recently established a set of reference data for photon beams from a particular linear accelerator, including test cases that are complete enough to enable extraction of characteristic input data for most available treatment planning systems.⁵⁵ This data set is primarily intended to serve as a test set to verify the performance of a treatment planning system for photon beams and should not be used in lieu of the actual beam data for the accelerator to be commissioned. AAPM Task Group 23 report is also a good overview of the kind of input data that might be needed for commissioning, the kind of equipment involved in their acquisition and the kind of test cases needed for quality assurance of a computerized treatment planning system for photons. In any case, the treatment planning system data should be modified such that the resultant isodose curves match the measured data.

Considering the many new types of questions that must be addressed in the clinical implementation of asymmetric collimators for photon radiation therapy, the AAPM has also formed a new task group (52) to address the dosimetric aspects of asymmetric collimators.⁵⁶

In the following subsections, the beam data necessary for square and rectangular x-ray fields, open and wedged, are described. This is followed by a more detailed description of dosimetry for blocked photon beams.

1. Square and rectangular photon beams

The following are needed for the calculation of the number of monitor units required to deliver a prescribed absorbed dose at a point at a given depth along the central ray of a square or rectangular beam in a unit density medium:

- (1) tables and/or graphs of percentage depth dose and/or tissue air ratios and/or tissue phantom ratios, for all square fields with suitable increments in dimensions;
- (2) a table of "equivalent square fields;"
- (3) a table of output factors in air and in phantom;
- (4) correction factors for changes in PDD for nonstandard SSDs;
- (5) peak scatter factors;
- (6) tray and wedge correction factors.

For the manual calculation of absorbed dose at an off-axis point, the following additional data are necessary:

- (1) isodose charts (for constant SSD) for square fields, with suitable increments in field size;
- (2) isodose charts (for constant SSD) for a selection of elongated fields, and/or suitable rules to convert charts for square fields to the desired rectangular field;
- (3) a method to correct for oblique incidence,

Measurements of PDD are usually taken for square fields of sides 4,5,6,8,10,12,15,20 and at further increments of 5 cm up to the largest setting. Field sizes are generally expressed in cm at the isocentric distance. Isodose charts are usually taken for the same square fields, and normalized at the d_{\max} on the central ray. These measurements are best taken in water using a scanning dosimetry system.

2. Wedged photon beams

Strictly, "wedge filters" are designed and applied to tilt the isodose lines, in one of the two principal directions, over a specified angle, while leaving the isodoses in the other direction essentially unchanged. The nominal angulations ("wedge angles") by which the wedges are indicated refer to the angle between the wedged isodose curve and the open beam isodose curve at a specified depth for a specified field size. Usually the chosen depth is 10 cm and the field size is 10X10 cm. The wedge angle changes very slowly with field size, but appreciably with depth, depending on the photon energy. The central axis depth dose data for wedged fields must be acquired because spectral changes in beam will cause a slight change in depth dose, particularly for large wedges such as 45° or 60°. Also, isodose curves must be measured for all wedged fields with a suitable increment for field size, usually 5 cm.

The wedge factor is defined as the ratio of the absorbed dose at a specified depth (some use d_{\max} and others use 10 cm), measured in the standard geometry with the wedge in place, to the absorbed dose in the same geometry without the

wedge. The dependence upon both field size and depth^{57,58} should be checked to see if a single factor is appropriate.

In most current designs there are two wedge positions for each wedge angle: one for each of two opposite directions (toe-in and toe-out). In some cases, wedges in the orthogonal direction are also available (toe-left and toe-right). Most often, the "zero direction" of the wedge slides is across the "zero direction" of the couch. For all wedges, it is important to perform frequent checks on:

- (1) mechanical play in the direction of the wedge slope;
- (2) the reliability of the locking mechanism securing the wedge in place; and
- (3) any difference in the output between the toe-in and toe-out positions or between the toe-left and toe-right positions.

If the maximum difference between the two or four wedge orientations is 1% or less, one may use a single averaged factor for all wedge orientations for a given field size and depth. Furthermore, if the maximum variation of the orientation-averaged values with depth and field size is 1% or less, a single overall wedge factor can be used.

It should be noted that some machines generate different wedge angles by mixing a 60°-wedged field and an open field with appropriate weights.

Recently, the ability to generate wedged fields using a moving collimator jaw during treatment (dynamic wedge) has become available. The task group considered a discussion of dynamic wedge beyond the scope of this document and refers the reader to the recent literature.⁵⁹⁻⁶²

3. Beam-shaping blocks for photons

Beam-shaping blocks are widely used to conform the basic rectangular field shape to the beam's eye view of the target volume. Its purpose is to function as auxiliary diaphragms, providing as much shielding of normal tissue as is reasonably possible, generally about 5%. In practice, the limiting conditions for blocks are their weight and thickness. The cross-sectional shape of shielding blocks is usually determined from the contours drawn on simulator x-ray films. Due to magnification, the actual size and shape of the cross section of a block is smaller than that shown in the simulator films. Customized blocks are cast from a high-density, low melting point alloy in Styrofoam forms, cut by a standard hot wire technique. The hot wire mimics a ray from the center of the source, tracing the prescribed contour on the x-ray film. Thus, the blocks are "focussed," which maximizes their effectiveness across their cross section by minimizing oblique transmission through the edges of the blocks. Nevertheless, in analogy to the penumbra of a regular field, there is a geometric as well as a physical penumbra effect around the edge of a shadow field produced by beam-shaping blocks.

The thickness of shielding blocks is a compromise between the desirability of a low transmission through the block and practical considerations of block size and weight. It should be noted that additional block thickness does not influence the amount of scatter received from the residual parent field (the unblocked portion of the original field). The effective transmission for a block is determined by the pri-

mary radiation component transmitted through the block and the scatter component from the collimators and the patient. This total scatter component, and therefore the effective transmission factor, depends both on the size of the parent field (i.e., the open field), including the effect of other blocks, the size of the shadow field and to some extent on its position in the parent field. Together, these considerations generally lead to a standard thickness of 4 to 5 half-value layers, or a “nominal” transmission factor, i.e., a reduction factor of the primary component of about 5%. Obviously, for a small shadow field in a large parent field the effective transmission can be considerably higher than the nominal transmission factor for the block.

For a parent (i.e., open) field, equivalent square $c \times c$, the block transmission factor (BTF) at depth d , for a centrally placed shadow (i.e., blocked) field, equivalent square $s \times s$ can be given in terms of the respective tissue-air-ratios (TAR), at a depth d , by:^{46,47,63}

$$\text{BTF}_{c,s}(d) = \left(1 - (1 - T) \frac{\text{TAR}_s(d)}{\text{TAR}_c(d)} \right), \quad (2)$$

where T is the nominal transmission factor, $\text{TAR}_s(d)$ is the tissue-air-ratio at depth d for (blocked) equivalent field $s \times s$; and $\text{TAR}_c(d)$ is the tissue-air-ratio at depth d for (open) equivalent field, $c \times c$.

In the case of multiple blocks, $c \times c$ should be taken as the equivalent square parent field with all blocks in place except the one presently considered. The present calculation method applies along the center line of the shadow field situated essentially within the “flat” part of the parent field. In the case of a laterally placed block, centered at some distance from the central ray, the off-center ratios of both the parent beam and shadow beam need to be taken into account.

Despite the common clinical application of shielding blocks, this subject has not been treated rigorously in most treatment planning systems. One approach is to simulate blocking by using negatively weighted beams and another is to employ a modified external contour in the treatment planning process.⁶⁴ Various commercially available treatment planning systems offer their own algorithms, often based on separation of primary and scattered radiation and subsequent Clarkson integration of scatter. These algorithms do not take into adequate account the penumbra around shadow fields. Size of penumbra governs the geometric margin between the optical shadow and the edge of effective shielding, which might be defined, for instance, as the line at which the effective attenuation is 90% of that at the center line of the shadow field; in other words, the 90% decrement line of effective attenuation. A simple calculational method that accounts for penumbra effects produced by shadow blocks is briefly described below.^{46,47,63}

(1) Measure central axis depth dose distributions in a number of clinically representative fields, in standard geometry, with and without centrally placed shielding blocks of standard thickness, casting shadow fields of various clinically representative sizes. These measurements are

instructive as to the mutual influence of parent field size and shadow field size on the effective attenuation versus the nominal attenuation.

(2) For the same geometric conditions, measure dose profile across the field, with and without the same blocks, at several depths. The full width at half maximum (FWHM) of the shadow profile should be equal to the projection of the block at that depth position. Let the full width at 90% of maximum shadow profile depth (FW90) be the width of acceptable effective shielding. For a given depth the margins, i.e. (FWHM-FW90), are, for practical purposes, independent of the parent field/shadow (i.e., open field/blocked field) field configuration. Also, the behavior of (FWHM-FW90) as a function of depth is primarily governed by simple geometry. These geometric properties should guide the radiation oncologist in prescribing the block contour on the planning film.

The above observations regarding the relationship between parent field and shadow field in determining the resulting dose distribution apply universally. Still, the special case of shielding small areas such as the lens of an eye deserve special attention because of the narrow tolerances required both in dose level and dose distribution across the lens.⁶⁵

It should also be noted that the output on the central axis of radiation can be affected by the presence of blocks even if the blocks do not project a shadow on the central axis.

D. Commissioning stationary electron beams

An extensive update on clinical dosimetry with electron beams was presented recently in the AAPM Task Group No. 25 Report on “Clinical Electron Dosimetry” (1990).⁶⁶ This report was, “primarily to fill the needs of a hospital physicist in the utilization of clinical electron beams of 5-25 MeV. Its scope has been restricted to (i) dosimetry measurement techniques and procedures for acquiring the basic information that is necessary for treatment planning and the acceptance testing of a new electron accelerator and (ii) the utilization of dosimetry data for the determination of monitor units. Principles of collimation and its influence on the patient dose are discussed for shields both external and internal to the patient, and the report states in an elementary manner the effect of tissue inhomogeneities on the dose distribution. Each section includes the procedure recommended by the task group for the performance measurements and calculations, the reasons for that choice, and a supporting bibliography.” In the present context attention will be limited to a listing of the basic data and methodology needed to use electron beams for actual patient treatment. For details the reader is referred to the Task Group No. 25 report.

Dose distributions in an electron field depend strongly on the design and construction of the collimating system, which may consist of a set of square field applicators (“cones”) providing basic square fields, or variable-jaw diaphragms providing continuously variable square and rectangular fields. In both cases, in the interest of sharp field definition and field flatness, the standard distance between diaphragm

and entry surface position (the "air gap") is usually small: on the order of 5 cm. Also, customized field shaping is very common in electron beam therapy. Because of these collimation systems and shields, it is usually necessary to treat some patients on slightly extended SSD to accommodate the cones. Although the electron beam is usually incident upon a flat skin surface, it is sometimes necessary to irradiate patients with electron beams entering the patient skin at an oblique angle. All of these variations in treatment conditions affect dose distributions produced in the patient. In addition, dose distributions produced by electron beams can be drastically affected by the presence of tissue heterogeneities in the irradiated volume. Although all of the above mentioned effects must be taken into account in clinical electron beam dosimetry, we recommend the following factors to be essential before releasing an accelerator for electron beam therapy:

- (1) calibration of beam output;
- (2) central-axis depth dose curves in water;
- (3) isodose charts in water;
- (4) cross beam profiles in water;
- (5) output factors;
- (6) corrections for field shaping; and
- (7) corrections for air gap.

The remaining issues dealing with effects of oblique incidence, patient contour, and tissue heterogeneities can be considered later, as needed.

Practical clinical electron dosimetry depends largely on calculational procedures to determine two- and three-dimensional distributions using computerized RTP systems. The potential users of any such system must pay close attention to its implications as to measuring equipment, data preparation, and verification of the dose calculation system before commissioning the accelerator for electron beams. Generally, commissioning electron beams needs much more time and effort than commissioning photons.

1. Dosimetry data for electron beams

With each cone, there is an optimum photon beam collimator setting recommended by the manufacturer for optimum electron beam characteristics, and the latest machines are automatically set for the optimum configurations. For commissioning electron beams it is necessary to measure the following beam characteristics for each cone separately:

- (1) output factor;
- (2) central axis depth dose curves;
- (3) isodose charts; and
- (4) cross beam profiles.

In the case of continuously variable diaphragms, it is recommended that the above mentioned beam characteristics be measured for an array of square fields of suitable sizes: at small increments with small fields, at larger increments for field sizes larger than the practical electron range.

These measurements should include all available electron energies and a series of elongated rectangular fields of sizes representative of clinically realistic shaped fields. The resulting information should be kept, in suitable graphic and tabular form, for clinical use.

Methods for determination of dose in electron beam Using ionization chambers, film, TLD, and diodes have been presented in the Task Group No. 25 report.⁶⁶ This report presents the data necessary for relative depth dose measurements not only in water but also in nonwater phantoms such as clear polystyrene, high-impact polystyrene, PMMA, and Solid Water. Thus, it is possible to measure dose distribution in these nonwater phantoms and subsequently convert these measurements to dosimetry in water using the Task Group No. 25 recommendations.⁶⁶

Dosimetry parameters depend strongly upon electron energy that must be specified for each clinical electron beam. Task Group No. 25 defines $E_{p,0}$ the most probable energy at phantom surface which is derived from the practical range, R_p , in water. It also defines the mean electron energy at the surface, E_0 , that is proportional to the depth of 50% ionization or dose, R_{50} . As the electron beam penetrates the phantom the mean electron energy at depth d , E_d , decreases linearly with depth, reaching a near zero value at a depth equal to the practical range, R_p .

The energy parameters $E_{p,0}$, E_0 , and E_d as well as range parameters R_p and R_{50} are useful for treatment planning. All of these parameters must be extracted from a direct measurement of a depth ionization or depth dose curve for every electron beam energy before releasing the machine for clinical use. From the same data, surface dose and x-ray contamination of each electron beam field should be determined. Also, the field size dependence of central axis depth dose should be evaluated as a function of electron beam energies. For each electron energy the size of electron field, below which loss of lateral electronic equilibrium influences the depth dose, should be determined.

From the isodose charts, the size of penumbra and its dependence on depth should be evaluated for each electron field. Guidelines about the field size necessary to cover a given target size should be developed, taking into account the constriction of isodose curves at shallower depths and their bulging at larger depths.

2. Field shaping for electrons

In clinical practice, most stationary electron fields are shaped by customized diaphragms cast in low melting-point alloy inserts placed in the regular cones. Often such odd-shaped, elongated and narrow fields are designed to "boost" the dose to superficial areas involving surgical scars in the wider target area. Both output and central ray depth dose distributions in the electron field may be significantly different from the open cone values.^{67, 68} For small field sizes, film dosimetry has been used effectively to obtain both output and central-axis dose distribution.^{69,70} Equivalent square rules for irregularly shaped electron fields are different from those for photon beams.⁷¹ Both the output factor and depth dose distribution are dependent on in-depth and lateral electron equilibrium.⁴⁴ As a safe rule of thumb, lateral buildup requires about one-half the dimension of the practical range.⁴⁴ Thus for fields where both dimensions are larger than the practical range, R_p , there will be lateral electron buildup, and d_{max} can be expected to be at its normal large-field depth. Conversely, if in a shaped electron field, the

smallest cross beam dimension becomes smaller than the practical range, the lateral as well as depth dose profile may be affected. In such cases individual measurements are necessary.

To determine the thickness of shielding material necessary for shaping fields, transmission curves through the shielding material for various electron energies should be measured. Such measurements in broad beams provide an upper limit to the shielding requirement for all field sizes. For internal shields such as eye shields, where minimum thickness is required, transmission measurements should be made specifically for the given field size and the depth of the structure to be shielded. Typical values of minimum thickness of lead needed for shielding electron beams can be obtained from the Task Group No. 25 report.

In the case of internal shielding it is also necessary to determine the contribution of electrons backscattered by the shield. This is particularly important for lower electron energies and shields made of higher atomic number materials. Again, the Task Group No. 25 report provides guidance on this matter.⁶⁶

3. Corrections for air gap or extended SSD

Electron beam therapy is usually given at a standard SSD with the patient skin surface at the isocenter for isocentrally mounted accelerators. If, for whatever reason, treatment needs to be given at other distances, corrections need to be made for changes in beam output as well as dose distribution. It is useful to create a library of typical isodose charts for some representative distributions, as well as central axis and cross beam dose/ionization profiles. Due to the changing collimator-skin geometry, both the surface (skin) dose and the buildup to the maximum depth dose may change relative to those for the standard distance.^{72,73} Also, beam flatness and penumbra are sensitive to the air gap size. The output factor requires inverse square correction relative to the effective source position. The latter should be derived from output measurements at several central ray positions, or from the values of the FWHM at various distances, using the isocenter as the reference position.⁷⁴

The methods that can be useful in making these corrections for air gap or extended SSD have been described in detail by Task Group No. 25 in its recent report.⁶⁶

4. Effects of oblique incidence and tissue heterogeneities

The electron beam dosimetry data discussed in the above subsections applies only to electron beams normally incident on water or water-equivalent homogeneous media with flat surfaces. In actual practice, the patient surface is usually curved and the patient has a considerable amount of tissue heterogeneities. If it is necessary to consider the effects of oblique incidence, patient contour, and tissue heterogeneities, many additional factors also need to be considered. For example, it is necessary to consider changes in electron scattering, beam penetration, and interface effects. Details about the effects of curved surfaces, oblique incidence, and tissue heterogeneities have been presented in the Task Group No. 25

report⁶⁶ Because of the evolving nature of this complex problem, its complete solution is not possible or is necessary for commissioning an electron beam. However, it is useful to make a few simple checks on the effects of oblique incidence. For example, the cross beam profile at a depth close to the surface for some angles of oblique incidence are instructive because a much narrower air gap may occur on one side of the beam than on the central axis air gap. With a narrow gap a "horn" may appear around the geometric edge of the field,⁶⁷ with a narrower than normal penumbra. On the other side of the obliquely incident beam, with the wider air gap, the profile may show a considerably rounded shoulder and a much wider penumbra.

V. COMMISSIONING OF SPECIAL PROCEDURES

A. Overview

The number of special procedures and new applications continues to grow as technology continues to provide a wider range of beam energies and more accurate, rapid and versatile control of treatment parameters. In this section, a number of now well-established special procedures are described. Because of their evolving nature, other important special procedures such as dynamic conformal therapy are omitted. The special procedures described here have all been subjects of interest to various AAPM task groups and the reader is referred to their reports for details. Only a brief discussion of these procedures, some of which are single treatment regimens with very little tolerance for error, is presented here for the sake of completeness.

B. Total and half body photon irradiation

A detailed discussion of total body and half body photon irradiation is contained in AAPM Report No. 17 (Ref. 75) and a recent review was presented at the 1990 AAPM Summer School.⁷⁶

Total body irradiation (TBI) has historically been associated with dedicated facilities specifically designed to provide large treatment fields.⁷⁷ Most modern facilities, however, use standard radiotherapy accelerators and achieve the large treatment fields for TBI or half body irradiation (HBI) by using either extended distances or moving beam techniques. These large fields and extended treatment distances require special consideration. The inverse square relationship which is valid in the vicinity of isocenter may not accurately predict the decrease in beam intensity at distances of 300 to 400 cm. This will then lead to errors in converting dosimetric quantities, such as percent depth dose and output, to the extended reference point. There may also be a change in photon spectrum between the small and large fields resulting in a corresponding change in mass energy absorption coefficient. At depth, the large fields also have an increased proportion of scatter as compared with smaller fields.⁷⁸ All data used at these extended SSDs should be carefully verified.

The large fields utilized in TBI also require large phantoms to provide full scatter for dosimetry measurements. Factors which correct for the loss of scatter if the phantom is not large enough to intercept the full beam have been re-

ported by Podgorsak⁷⁹ and Van Dyk.⁸⁰ If compensation is required, special broad beam attenuation data should be used.⁸¹

C. Total skin electron irradiation

An extensive discussion of treatment options, dosimetry techniques, instrumentation, and the consideration of various patient parameters is given in the AAPM Report No. 23 entitled "Total Skin Electron Therapy: Technique and Dosimetry"⁸² and a recent review presented at the 1990 AAPM Summer School.⁸³ Only a brief discussion is presented below.

The treatment of diseases such as cutaneous T-cell lymphoma often involves the treatment of the entire body surface to a depth of at most 1 cm. The use of total skin electron therapy can often provide an effective dose to these tissues. Acceptable treatment techniques must provide for large treatment fields, adequate penetration, sufficient dose rate, field uniformity, low x-ray background and procedures to evaluate and boost underdosed regions as well as to shield high dose regions of sensitive tissues. The most common technique employs six angulated beams of low energy electrons (4-10 MeV) at extended source to surface distances of approximately 3 m. The minimum necessary field size is 200 cm in height and 80 cm in width at the patient plane. It is common to employ six dual fields (12 fields) to cover the entire skin when extended SSD availability is limited.

Increased beam currents are often necessary to maintain the desired dose rate at these extended sources. While the increased load on the accelerating structure falls well below the level used for x-ray production, a careful evaluation of the dosimetry system must be carried out. The linearity of the monitor chambers and their associated electronics must be verified. Also, collection efficiency of monitor ion chambers should be checked. Film and TLD dosimetry can be helpful in obtaining dose contribution due to multiple fields.⁸⁴ Consideration should also be given to the additional load placed upon other accelerator components by the increased dose rate necessary for total skin irradiation.

The six dual field technique poses significant dosimetric challenges.⁸² These include single beam calibration, the determination of the shift in percent depth dose and maximum dose resulting from the oblique entry of adjacent beams, the determination of the average surface dose, evaluation field uniformity over an electron field size of 80X200 cm and measuring the photon background which will penetrate past the prescribed treatment depth and be cumulative for all fields treated. For details, the reader is referred to AAPM Report No. 23.⁸²

D. Electron arc therapy

Single electron field treatment of shallow targets extending over large body contours can result in significant dose inhomogeneities in the target volume. These inhomogeneities become more pronounced when the surface contour contains slopes of 30° or more. If such targets are treated with several smaller fields, a choice between increased or decreased dose along the lines of field abutment must be made. An alterna-

tive treatment technique using arcing electron beams can provide a more uniform dose throughout the target region. Electron arc therapy is, however, a difficult technique to execute, having many unique problems to overcome, including generation of accurate treatment plans and computation of MU/deg values.

To shape the electron fields properly and to compensate for dose inhomogeneities, three levels of collimation are required; primary x-ray collimation, secondary collimation for electron slit shaping and tertiary collimation on patient skin. The setting of the x-ray jaws must take the electron dose as well as the x-ray background into account.⁸⁵⁻⁸⁷ The tertiary collimators are used not only to confine the electron beam but also to provide a means of compensating for the increase in output with a decrease in radius of curvature.^{85,88-90} The fourth level of collimation, which is placed in contact with the patient's surface, restricts the beam to the target volume and sharpens the penumbra.^{85,91}

The therapeutic range of the effective beam achieved by electron arc therapy is greater than for the same beam used in a static mode. This is accompanied by a decrease in surface dose⁹²⁻⁹² which usually dictates the need for bolus. As a result of the constant focus of the beam at an isocentric point, x-ray doses approaching 4% for a 6 MeV beam to 26% for an 18 MeV beam have been reported⁹³⁺⁹⁵

The output for electron arc therapy has been defined as "the maximum dose per monitor unit in a cylindrical water phantom on central axis which is the axis passing through isocenter and bisecting the arc in the central plane of rotation."** For a given treatment, the dose per monitor unit, radius of curvature of the skin surface, tertiary collimator width, depth of dose maximum source to axis distance, source to surface distance, as well as beam energy must be carefully considered in treatment planning.

The dose measurement technique for arc electron beams has been reported by several investigators.⁸⁵⁻¹⁰¹ Leavitt has provided a recent review of the electron arc therapy technique at the 1990 AAPM Summer School,¹⁰¹ and the AAPM Task Group No. 37 has detailed recommendations on electron arc therapy.¹⁰²

E. Intraoperative radiotherapy

Intraoperative radiotherapy (IORT) is a multidisciplinary procedure which combines two conventional methods of cancer treatment, namely, surgery and radiation therapy. The purpose is to deliver a large single dose to the surgically exposed tumor bed while minimizing dose to normal tissues.^{103,104} In the U.S. IORT is generally an adjuvant therapy, i.e., that it is given as a boost after conventional, fractionated radiotherapy.

One of the unique features of IORT is the requirement to deliver the radiation into a sterile operating field. Another is the wide range of technical and physical information necessary at the time of the procedure. For each available energy, the isodose distribution as well as the relative output for each cone must be known and readily available to the medical team in the operating room. The effects of field shaping with lead sheets must also be well understood and documented.

While orthovoltage units have been used for IORT,¹⁰⁵ the most frequently used treatment modality is the electron beam from a linear accelerator. The use of special sterile cones, which the surgeon and the radiation oncologist insert and secure in place during surgery, is necessary. Each cone must be designed to enable the physician to view the area defined by the end of the cones. Cones usually encompass rectangular and circular shapes with various beveled angles. The change in percent depth dose associated with these cones as compared to routine electron applicators has been documented¹⁰⁶ but must be measured for an institution's cones. The variation of therapeutic range along the central axis of the beam versus that perpendicular to the end of a beveled applicator has also been discussed.^{72,106} The design, shape, and size of each cone affects its depth dose, x-ray contamination, dose output calibration, flatness and symmetry, leakage through the cone walls and shape of the dose distribution.¹⁰⁴

Cone systems usually contain two sections. The distal portion, which is aligned in the patient and the mating section, which is secured to the linear accelerator. Cone designs can be broken down into two types: docking^{103,106-110} and nondocking.^{104,111,112} The wide range of cone shapes and sizes as well as the electron energies used in operation have been documented.¹¹³ For both docking types, care must be taken to shield against leakage both at the points of cone connection and through the cone wall. While leaded acrylic wall thickness of 3 mm has been reported as effective against leakage,¹¹³ several reports recommend acrylic wall thicknesses of approximately 6 mm (1/4").^{103,104} For either docking type the distal and proximal ends of the cone must be brought into alignment. Because of limited motion of a treatment head on isocentrically mounted gantries on modern accelerators, special IORT tables are often necessary.¹⁰³

As with any electron applicator, any material placed in the path of the electron beam, such as a thin mylar window to isolate the accelerator from the patient, will affect the cone calibration.¹¹³

Most adjustments to flatness and symmetry are affected by varying the x-ray jaw settings.^{103,107,110} An alternative approach is to introduce a steel ring onto the inner surface at the lower edge of the beam that protrudes onto the beam. This helps to reduce the dose at the edge of the field, allowing for better homogeneity across the cone's opening.¹⁰⁴

The dosimetric data for IORT include the basic parameters of routine electron beam therapy. These include photon collimator setting, applicator cone ratio, central axis percent depth dose, surface dose and buildup, x-ray contamination, and cross-field behavior. The added concerns of dose verification and the effects of any field shaping must be taken into account. A discussion of the necessary dosimetric measurements is given in the 1986 AAPM Summer School¹¹⁴ and the 1990 AAPM Summer School proceedings.¹¹⁴ Also, a task group (No. 48) of the AAPM is currently developing guidelines for IORT.¹¹⁵

F. Stereotactic radiosurgery

The term stereotactic radiosurgery denotes an external beam technique which utilizes high precision localization

and delivers a large dose in a single fraction for the treatment of intracranial targets. The intent of the treatment is to cause necrosis in the target volume using radiation. More recently, this technique has been extended to treat tumors using multiple fractions and doses used in radiotherapy. Execution of the stereotactic procedure involves the use of an extra-cranial reference system which has been fixed relative to the patient's anatomy. Many of the stereotactic frames fix directly to the skull by means of pins or anchor posts^{116,117} employed where repeat fixation is required.¹¹⁸ Several new stereotactic frames have recently been suggested, which are located relative to the bony anatomy and are acceptable for repeat application, thus allowing fractionated therapy.^{119,120}

Each stereotactic reference system has associated accessories for angiography, computed tomography and magnetic resonance imaging. These localization systems allow the position of an intracranial target to be calculated relative to the reference coordinates of the stereotactic fixation frame. The localization devices superimpose reference points onto the diagnostic images. This allows the position of any point in the image to be calculated relative to the stereotactic reference frame.^{121,122}

After target localization has been accomplished, the application of either noncoplanar arcs or multiple stationary beams is planned. The multiarc techniques allow high dose gradients to be obtained in all directions around the target volume. Typical beam diameters range from 4 to 40 mm. Most common target diameters for the treatment of arteriovenous malformations are found to be between 24 and 30 mm.¹²³ For targets which depart from spherical geometry, multiple isocentric plans may provide the best therapeutic ratio. More recently, conformal therapy has been suggested for those targets that significantly depart from spherical geometry.¹²⁴

The overall accuracy is a combination of treatment accuracy and localization accuracy. For vascular targets, the primary imaging modality is biplanar angiography. While angiography will allow the reconstruction of a single point to a precision of less than 0.1 mm, the errors inherent in clinical delineation of target size and shape can lead to errors in the estimation of target boundaries in excess of 5 mm. For CT and MR localization, scan diameters on the order of 30-35 cm are often required to image the anatomy and stereotactic localizer frame. For 512X512 pixel images, this results in an individual pixel dimension on the order of 0.7 mm with routine slice thicknesses of 1.5 to 5.0 mm.

The accuracy achievable in "delivering" the treatment depends upon the gantry isocentricity and patient rotational accuracies for accelerator-based systems. The careful documentation of all potential errors throughout an individual system will allow their simulation with a treatment planning system and the subsequent determination of the margin necessary to guarantee target coverage in spite of the uncertainties.

Stereotactic radiosurgery has five qualities which distinguish it from more routine treatment techniques. These are: (1) determination of the target volume cannot be done using routine therapeutic simulation techniques; (2) three-dimensional treatment planning is always necessary in this

procedure; (3) more stringent isocenter criteria for gantry and patient motions are required; (4) small beam sizes are utilized; and (5) the treatment is often delivered in a single fraction. For more stringent requirements on the accuracy and relative novelty of stereotactic radiosurgery, we present a more detailed discussion of this technique in the following subsections.

1. Linearity of localization images

For diagnostic imaging, spatial resolution and linearity are often sacrificed in exchange for increased contrast. An example of tradeoff is digital angiography of vascular targets. The gains in image definition in this digital angiographic procedure usually are at the expense of spatial linearity within the entire field of view. For magnetic resonance images, local field perturbations can result in local spatial errors within the reconstructed image. Variations in spatial accuracy of MR images may also depend upon the plane of reconstruction.¹²⁵ To avoid these errors, careful analysis of all imaging modalities to be used with the procedure must be carried out and spatial correction must be applied.

Some localizing systems use data not contained within the image, such as CT couch position and gantry angle, or require precise alignment, such as image orthogonality for bipolar angiography. When these systems are used, the diagnostic imaging equipment must be subjected to the same basic quality assurance program as the therapy equipment. For an angiographic localizer, accuracy of orthogonality and spacing of fiducial points as well as projected source positions must be assured for each procedure. For CT and MR localizers, similar checks on integrity of the localizer should also be incorporated into the image acquisition algorithms. Aside from utilizing redundant information, known phantom target points can be placed into each image. The algorithm can then reconstruct this target as an internal consistency check.

2. Treatment planning

To achieve the desired steep dose gradients, it is essential to use multiple beams from different directions toward the target. Usually it is necessary to use many noncoplanar beam paths. To evaluate dose distributions produced by such a complex three-dimensional configuration of beams, proper planning systems must be capable of true three-dimensional computations. The display of the dose distribution in axial, coronal, and sagittal planes is not only desirable, but necessary to arrive at the most appropriate treatment plan.

The patient data set needed for treatment planning includes the images used for target localization and a set of scans, most commonly axial, which extend through target regions and which include all surfaces through which beams will enter. The system should allow the display of the dose distributions anywhere within the skull. While dose delivered at a volume far away from the target region may be small, it is important to evaluate the entire irradiated volume if for no other reason than to inspect for inadvertently overlapping beams. The planning system must also be able to compute the dose along finely spaced grids. Dose gradients on the

order of 10% per millimeter are not uncommon. A 1X1 mm grid, or smaller, in the vicinity of these high dose gradients is considered essential.

Because most commercial treatment planning systems do not fully support this procedure, locally developed systems are commonplace. If these systems are to be employed, adherence to good QA testing must be strictly enforced. The only computer code which should be used for patient treatment is a fully-tested code. If the code is altered, even to correct an error, the code must be fully tested again. Because of the complex nature of the localization and treatment procedure even large errors in localization and dose computation can go unnoticed.

As in any treatment planning process, one of the primary tasks of planning a radiosurgical treatment is to avoid critical normal tissues. It is therefore very important for the radiation oncologist and the neurosurgeon to be able to identify all areas of intracranial function during the planning process. This can lead to technical difficulties if the planning process depends upon distinctly contouring each area of silent and nonsilent brain function since many areas are not separated by distinct tissue planes. While the contouring can yield valuable surface-rendered views it has been suggested that this approach compresses, and at times deletes, very valuable information which is available in the original CT and MRI images.^{126,127} It has been suggested, by the same authors, that during the planning process reformatted CT and MR imaging should always be available. This will ensure that the information which is compressed or deleted during the contouring process is available during plan evaluation.

3. Dosimetry

One of the most difficult areas of the stereotactic radiosurgical procedure is the ability to measure the absolute dose as well as the relative dose distributions from the small fields employed. Small volume ion chambers, diodes, thermoluminescent dosimeters, and film have all been used for both relative and absolute dosimetry.¹²⁸⁻¹³⁶ The beam diameter is usually smaller than the sensitive volume of a standard Farmer type ion chamber. Ion chamber measurement of these small beams and steep dose gradients often suffer from lack of lateral equilibrium and chamber volume effects. Techniques for correcting these errors have been reported.¹³⁷ Due to the general uncertainty and difficulty in these measurements, measured data should be verified using two or more dosimetric methods.

4. Treatment unit

Several patient alignment systems for accelerator-based units have been developed. These include systems involving multiple stationary gantry positions, patient rotations,¹²⁷ arcing gantry, multiple fixed patient rotational positions, dynamic gantry, and patient movements.^{127,132,137-140} A comparison of these techniques has also been reported.¹⁴¹

Most systems employ tertiary collimators to provide steeper dose gradients. This places the final collimation at approximately 30 cm from the isocenter. The close proximity of the collimator system coupled with the complex motion

used in treatment presents a larger potential for collimator/patient collisions than conventional accelerators. Therefore, particular attention should be taken to assure the safety of patients from the risk of collisions, especially for computer-controlled accelerators.

5. Quality assurance of stereotactic radiation treatment

It is essential that in any of the above systems a comprehensive quality assurance procedure defines the alignment of the apparatus throughout the potential therapeutic volume. If, for example, the defined stereotactic volume in which a target may be positioned is $10 \times 10 \times 15 \text{ cm}^3$, then the accuracy throughout this entire matrix should be thoroughly tested. If the unit is disassembled and reassembled between uses, a procedure to validate the correct realignment should be executed prior to patient treatment. One of the first techniques suggested for measurement of an accelerator-based system's overall accuracy involves the placement of a small spherical target at the isocenter. Portal films are taken, using the treatment beam, at gantry and patient orientations which are representative of the range of motion used in treatment. The movement of the target relative to the beam edges provides a measure of the total accuracy of the beam delivery system. If the procedure is repeated at a set of isocentric coordinates through the potential therapeutic volume, an overall assessment of beam accuracy can be established. By placing the target at the isocentric coordinates to be used for a specific patient, the overall accuracy of an individual treatment can be measured. It is recommended that this or an equivalent test should precede each treatment. This not only validates the reassembly and alignment of the stereotactic radiosurgery treatment system but can also ensure the correct setting of the system for an individual patient treatment.

Since most stereotactic radiosurgery procedures involve single (or very few) treatment events, a system of double checks at each phase of data entry and treatment setup should be employed. This procedure should incorporate different individuals to minimize human error. A review of stereotactic radiosurgery is available in the 1990 AAPM summer school proceedings¹⁴² and an AAPM Task Group (No. 42) is almost ready to present its recommendations on this matter.¹⁴³

VI. QUALITY ASSURANCE PROGRAMS

A. Overview

Several quality assurance programs^{20,21,25,27,28} are needed to ensure the safe and efficacious application of radiation for treatment of cancer. Major components of the quality assurance programs deal with radiation protection of personnel and patients, safe maintenance and operation of the accelerator and accuracy of dose delivery to the correct target volume. These programs are described below.

B. Radiation protection of personnel and patients

Film badges are required for all personnel who work frequently in the vicinity of radiotherapy accelerators. This list

includes, but is not limited to, radiation therapists, physicists, physicians, dosimetrists, accelerator maintenance personnel, nurses, and aides. An additional film badge should be placed in the vicinity of the console to monitor the exposure at that locality.

Daily, monthly, and annual radiation protection checks need to be carried out on both the machine and the facility. Checks on the machine include beam stopper interlock (if appropriate). Checks on the facility include the door interlocks (and search button, if one exists). Some of these checks have already been described in Sec. II.

C. Safe maintenance and operation of machine

A physicist must be responsible for accelerator operational issues such as safety, security, maintenance, QA, training, etc. In many cases, a physicist will coordinate these activities with other groups (security, biomedical engineering, etc.) in the hospital and/or clinic. Some of these activities are illustrated below.

1. Safety

Emergency instructions (perhaps different sets, operator-in-the-mom, and operator-outside-the-room) should be formulated and posted. The staff must be made familiar with the contents of the emergency instructions (these instructions might include procedures to follow in case of fire, or of certain types of mechanical failure, or of flooding, etc.). Hospital (and/or clinic) safety and security personnel should receive training in accelerator safety issues. For example, the institution's safety and security staff should know where the main circuit breaker for the accelerator is located.

2. Security

It must be assured that only authorized personnel can operate the accelerator. The requisite security measures might include requiring radiation therapists to carry the console key with them whenever the console is unattended and to lock up the console key(s) after hours. The institution's security staff should be instructed as to "normal" after hours activity at the accelerator.

3. Maintenance

A physicist should be responsible for obtaining machine maintenance and for assuring that the accelerator is fit to use after maintenance (be it a repair or "just" preventative maintenance). Regardless of whether the manufacturer's representatives or an in-house maintenance group have done the work on the accelerator, the physicist must be aware of what was done and how it might affect the accelerator's operation. For example, if repairs involved dosimetry components, the physicist must check the machine's output calibration (and perhaps flatness, symmetry, percent depth dose, and percent depth ionization) before allowing the accelerator back into clinical service. A service log book should be kept for each accelerator.

4. Training

After the manufacturer's initial support, a physicist is responsible for training as regards to safety features. He/she is a member of a team that usually includes the chief radiation therapist, as well as vendor applications staff. This training may be as complex as formal instructional sessions that include emergency procedures as well as accelerator operation. It will require that the user be very familiar with the operator's instruction manual for the accelerator. In any case, the end-point should be fully qualified operators. A record of training and a list of qualified operators should be maintained.

5. Logistics

The physicist may participate in safety efforts that include such logistical items as the preparation of appropriate forms (e.g., daily output check, etc.), proper location of the accelerator instruction manuals, proper location of emergency instructions, proper location of fire extinguishers, proper emergency lighting, proper voice and visual communication between the patient and radiation therapists, etc.

D. Accuracy of dose delivery

An effective quality control program to prevent treatment errors due to machine malfunction is essential for the safe and accurate delivery of radiation therapy. The program should be jointly established and implemented by radiation oncologists, physicists, engineers, and therapists. The basic components for such a program include well defined criteria, adequate instrumentation, qualified personnel, and sufficient documentation. The program should be simple to implement, but sufficiently comprehensive in scope. A record of quality assurance checks should be maintained and periodically reviewed for compliance.

In addition to the initial acceptance of the R & V system, an ongoing QA procedure should be established by the medical physicist to assure the correct transfer of information from the record and verify system to the accelerator, at least for each new field setting made.

1. Frequency of tests

The performance of a radiotherapy accelerator should be tested after machine repair or modification, at least in areas which may be affected by the repair. In addition, certain tests should be performed at regular intervals. The frequency of various tests should inversely correlate with the stability of the respective parameters tested, and should be based on experience with the equipment. Critical parameters, such as machine output and laser accuracy, should be checked at intervals that are short compared to a typical treatment course of 4-6 weeks. The frequencies of tests are guidelines and may be adjusted judiciously based on established records of individual therapy units.

2. Guidelines for quality assurance tests, tolerances, and frequencies

Recommended QA tests, their frequencies and criteria of acceptance are described in AAPM Report No. 13.²¹ An update of these recommendations is almost ready to be presented by AAPM Task Group No. 40.²⁸

3. Levels of alert

Appropriate corrective actions should be taken when specified tolerances are exceeded. Depending on the nature and seriousness of the malfunction, and the complexity of the remedy, the corrective action may be immediate or delayed. In all cases, the observation, the decision and/or corrective action should be documented, and where appropriate, clearly communicated to personnel whose work function would be affected.

4. Documentation and reports

The quality assurance procedures should be documented in detail. Data from quality assurance tests must be recorded in archival form. Where appropriate, data may be recorded on forms customized for the required periodic checks for specific machines.

All records of quality assurance should be reviewed on a regular basis by a qualified physicist. This is particularly important when some procedures are carried out under the supervision of qualified experts in other areas.

QA results should be summarized periodically and reviewed by a Quality Assurance Committee.

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APPENDIX A: DEFINITION OF A QUALIFIED MEDICAL PHYSICIST

A qualified medical physicist is an individual who is competent to practice independently one or more of the subfields of medical physics.

1. Therapeutic radiological physics

This particular field pertains to:

- (1) the therapeutic applications of x rays, of gamma rays, of electron and charged particle beams, of neutrons, and of radiations from sealed radionuclide sources;
- (2) the equipment associated with their production, use, measurement, and evaluation;
- (3) medical health physics associated with this subfield.

2. Diagnostic radiological physics

This particular field pertains to:

- (1) the diagnostic applications of x rays, of gamma rays from sealed sources, of ultrasonic radiation, of radio-frequency radiation, of magnetic fields;

- (2) the equipment associated with their production, use, measurement, and evaluation;
- (3) the quality of the diagnostic image resulting from their production and use;
- (4) medical health physics associated with this subfield.

3. Medical nuclear physics

This particular field pertains to:

- (1) the therapeutic and diagnostic applications of radionuclides (except those used in sealed sources for therapeutic purposes);
- (2) the equipment associated with their production, uses measurement, and evaluation;
- (3) medical health physics associated with this subfield.

4. Medical Health Physics

This particular field pertains to:

- (1) the safe use of x rays, of gamma rays, of electron and other charged particle beams, of neutrons, of radionuclides, and of radiation from sealed radionuclide sources for both diagnostic and therapeutic purposes, except, with regard to the application of radiation to patients for diagnostic or therapeutic purposes;
- (2) the instrumentation required to perform appropriate radiation surveys.

It is expected that an individual will not hold himself/herself out to be qualified in a subfield for which he/she has not established competency. An individual will be considered competent to practice one or more of the subfields of medical physics if that individual is certified in that subfield by any of the following organizations:

- (a) The American Board of Radiology,
- (b) The American Board of Medical Physics,
- (c) The American Board of Health Physics,
- (d) The American Board of Science in Nuclear Medicine,
- (e) The Canadian College of Physicists in Medicine.

The American Association of Physicists in Medicine regards board certification, in the appropriate medical physics subfield, and state licensure, in those states in which licensure exists, as the appropriate qualification for the designation of qualified medical physicist.

In addition to the above qualifications, a qualified medical physicist shall meet and uphold the "Guidelines for Ethical Practice for Medical Physicists" as published by the American Association of Physicists in Medicine.

APPENDIX B: RESPONSIBILITIES OF A RADIATION ONCOLOGY PHYSICIST

- (1) To develop requirements and specifications for the purchase of appropriate equipment.
- (2) To plan the facilities to house the accelerator (including shielding design).
- (3) To participate in, oversee, and monitor facility construction as needed.

- (4) To monitor machine installation by the manufacturer and provide assistance as needed.
- (5) To perform acceptance testing of the machine after installation.
- (6) To commission the machine for clinical use.
- (7) To establish methods for special clinical procedures and to acquire the necessary dosimetry data for them. These include special eye blocks, breast setups, total crania-spinal irradiation setup, electron arc, intraoperative electrons, total skin irradiation, total body irradiation, stereotactic radiotherapy, etc.
- (8) To establish procedures for monitor unit calculations for the accelerator.
- (9) To establish methods for the determination of dose distributions in the patient irradiated by the accelerator.
- (10) To participate in patient data acquisition, treatment planning and implementation, and evaluation of radiation treatments using the accelerator.
- (11) To implement and monitor a quality assurance program for personnel safety.
- (12) To implement and monitor a quality assurance program for patient safety and accuracy of dose delivery.
- (13) To implement and monitor a maintenance schedule for the accelerator.
- (14) To develop new procedures which may lead to better and more cost-effective use of accelerators in radiation oncology.

APPENDIX C: INSTRUMENTATION NEEDED FOR ACCEPTANCE TESTING AND COMMISSIONING OF A RADIOTHERAPY ACCELERATOR

Ionization chamber dosimetry system: two ionization chambers, two electrometers, constancy checkers, cables, thermometer, barometer, phantoms.

Film dosimetry system: densitometer, phantoms.

TLD dosimetry system: reader, ovens, jigs, phantoms.

Dosimetry scanning system: electrometers, scanning devices.

Personal computer system: computer, software for report generation, data collection and analysis, printer and plotter.

Quality assurance devices: survey meters, area monitors, jigs.

Beam modifiers: fabrication tools including hot-wire cutter for block fabrication.

¹P. Wootton, P. R. Almond, J. G. Holt, D. B. Hughes, D. Jones, C. J. Karzmark, and R. J. Schulz, "Code of practice for x-ray therapy linear accelerators," *Med. Phys.* 2, 110-121 (1975).

²(ICRU), Determination of absorbed dose in a patient irradiated by beams of x or gamma rays, ICRU Report No. 24 (International Commission on Radiation Units and Measurements, Bethesda, MD, 1976).

³E. J. Joyce, "Malfunction 54: Unraveling deadly medical mystery of computerized accelerator gone awry," *Am. Med. News* 1, 3, Oct. (1986).

⁴E. J. Joyce, "Firm warns of another Therac 20 problem," *Am. Med. News* 20, 7, Nov. (1986).

⁵E. J. Joyce, "Software 'bug' discovered in second linear accelerator," *Am. Med. News* 20, 7, Nov. (1986).

⁶E. J. Joyce, "Accelerator linked to 5th radiation overdose," *Am. Med. News* 1, 6 Feb. (1987).

⁷E. J. Joyce, "Software flaw known before radiation killed 2," *Am. Med. News* 3, 16, Jan. (1987).

- ⁸C. J. Karzmark, "Procedural and operator error aspects of radiation accidents in radiotherapy," *Int. J. Rad. Oncol. Biol. Phys.* 13, 1599-1602 (1987).
- ⁹Inter-Society Council for Radiation Oncology report, *Criteria for Radiation Oncology in Multidisciplinary Cancer Management* (American College of Radiology, Reston, VA, 1991).
- ¹⁰P. Biggs, "An interactive computer graphics program for determining the shielding for a megavoltage radiotherapy facility," *Health Phys.* 43, 601-607 (1982).
- ¹¹National Council on Radiation Protection and Measurements (NCRP), *Radiation Protection Design Guidelines for 0.1-100 MeV Particle Accelerator Facilities*, NCRP Report No. 51 (National Council on Radiation Protection and Measurements, Bethesda, MD, 1977).
- ¹²National Council on Radiation Protection and Measurements (NCRP), *Neutron Contamination from Medical Accelerators*, NCRP Report No. 79 (National Council on Radiation Protection and Measurements, Bethesda, MD, 1984).
- ¹³P. J. Biggs, "Shielding requirements for high energy accelerators," in *Advances in Radiation Oncology Physics*, edited by J. Purdy (American Institute of Physics, New York, 1991).
- ¹⁴American Association of Physicists in Medicine (AAPM), *Neutron Measurements Around High Energy X-Ray Therapy Machines*, AAPM Report No. 19 (American Institute of Physics, New York, 1987).
- ¹⁵R. E. Apfel, B. T. Chu, and J. Mengel, "Superheated drop nucleation for neutron detection," *Appl. Sci. Res.* 38, 117-122 (1982).
- ¹⁶R. E. Apfel and Y. C. Lo, "Practical neutron dosimetry with superheated drops," *Health Phys.* 56, 79-83 (1989).
- ¹⁷N.E. Ipe D. D. Busick, and R. W. Pollack, "Factors affecting the response of the bubble detector BD-100 and a comparison of its response to CR-39," *Rad. Protect. Dos.* 23, 135-138 (1988).
- ¹⁸N. E. Ipe, R. J. Donahue, and D. D. Busick, "The active personnel dosimeter-Apfel Enterprises superheated drop "detector," *Proc. Solid State Dosimetry Conference*, Vienna, 1989.
- ¹⁹National Council on Radiation Protection and Measurements (NCRP), *Recommendations on Limits for Exposure to Ionizing Radiation*, NCRP Report No. 91 (National Council on Radiation Protection and Measurements, Bethesda, MD, 1987).
- ²⁰E. Hanks, "Quality assurance in radiation therapy: Clinical and physical aspects," in *Proceedings of the First International Symposium on Quality Assessment in Radiation Oncology* (Pergamon, Washington, DC, 1983).
- ²¹American Association of Physics in Medicine (AAPM), *Physical Aspects of Quality Assurance in Radiation Therapy*, AAPM Report No. 13 (American Institute of Physics, New York, 1984).
- ²²International Electrotechnical Commission (IEC), *Medical Electrical Equipment: Medical Electron Accelerators; Section Four: Functional Performance Characteristics*, IEC Report No. 976, 35-II Sub-Committee No. 62C of Technical Committee No. 62 (International Electrotechnical Commission, Vienna, 1985).
- ²³International Electrotechnical Commission (IEC), *Medical Electrical Equipment: Medical Electron Accelerators: Guidelines for Functional Performance Characteristics* IEC Report No. 977, 36-I and II Sub-Committee No. 62C of Technical Committee No. 62 (International Electrotechnical Commission, Vienna, 1985).
- ²⁴Institute of Physical Science in Medicine (IPSM), *Commissioning and Quality Assurance of Linear Accelerators*, ISPM Report No. 54 (Institute of Physical Science in Medicine, York, United Kingdom, 1988).
- ²⁵World Health Organization (WHO), *Quality Assurance in Radiotherapy* (World Health Organization, Geneva, Switzerland, 1988).
- ²⁶M. S. Weinhaus, J. A. Purdy, and C. O. Granda, "Testing of a medical linear accelerator's computer control system," *Med. Phys.* 17, 95-102 (1990).
- ²⁷G. Kutcher, "Comprehensive quality assurance," in *Advances in Radiation Oncology Physics*, edited by J. A. Purdy (American Institute of Physics, New York, 1991).
- ²⁸G. Kutcher, Chairman, Task Group No. 40. *Comprehensive Quality Assurance for Radiation Oncology, Radiation Therapy Committee* (American Association of Physicists in Medicine, College Park, MD), *Med. Phys.* 21, 581-618 (1994).
- ²⁹J.A. Purdy, Chairman, Task Group 35, *Accelerator Safety, Radiation Therapy Committee* (American Association of Physicists in Medicine, College Park, MD, in progress).
- ³⁰F. J. Bova and L. T. Fitzgerald, "Errors in isocenter of Picker C-9000 teltherapeutic unit," *Health Phys.* 44, 553-554 (1983).
- ³¹R. J. Schulz, P. R. Almond, J. R. Cunningham, J. G. Holt, R. Loevinger, N. Suntharalingam, K. A. Wright, R. Nath, and G. D. Lempert, "A protocol for the determination of absorbed dose from high-energy photon and electron beams," *Med. Phys.* 10, 741-771 (1983).
- ³²R. J. Schulz, "Reply to comments of Wu et al.," *Med. Phys.* 11, 872-874 (1984).
- ³³R. J. Schulz, P. R. Almond, G. Kutcher, R. Loevinger, R. Nath, D. W. O Rogers, N. Suntharalingam, K. A. Wright, and F. Khan, "Clarification of the AAPM Task Group 21 dosimetry protocol," *Med. Phys.* 13, 755-759 (1986).
- ³⁴D. W. O Rogers, "Fundamentals of high energy x-ray and electron dosimetry protocols," in *Advances in Radiation Oncology Physics*, edited by J. A. Purdy (American Institute of Physics, New York, 1991).
- ³⁵International Atomic Energy Agency (IAEA), *Absorbed Dose Determination in Photon and Electron Beams. An International Code of Practice*. Technical Report Series No. 277 (International Atomic Energy Agency, Vienna, 1987).
- ³⁶L. Horton, *Handbook of Radiation Therapy Physics* (Prentice-Hall, Englewood Cliffs, NJ, 1987). 1st ed.
- ³⁷C. G. Orton and J. B. Seibert, "The measurement of teletherapy unit timer errors," *Phys. Med. Biol.* 17, 198-205 (1972).
- ³⁸F. H. Attix, *Introduction to Radiological Physics and Radiation Dosimetry* (Wiley, New York, 1986), 1st ed.
- ³⁹M. Weinhaus and J. Meli, "Determining P_{sc} , the correction factor for recombination losses in an ionization chamber," *Med. Phys.* 11, 846-849 (1984).
- ⁴⁰J. Dawson, J. M. Harper, and A. C. Akinradewo, "Analysis of physical parameters associated with the measurement of high energy x-ray penumbra," *Med. Phys.* 11, 491-497 (1984).
- ⁴¹D. J. Dawson, N. J. Schroeder, and J. D. Hoya, "Penumbra measurements in water for high energy x-rays," *Med. Phys.* 13, 101-104 (1986).
- ⁴²R. K. Rice, J. L. Hansen, G. K. Svensson, and R. L. Siddon, "Measurements of dose distributions in small beams of 6 MV x-rays," *Phys. Med. Biol.* 32, 1087-1099 (1987).
- ⁴³D. E. Velkley, D. J. Manson, J. A. Purdy, and G. D. Oliver, Jr. "Build-up region of megavoltage photon radiation sources," *Med. Phys.* 2, 14-19 (1975).
- ⁴⁴H. E. Johns and J. R. Cunningham, *The Physics of Radiology* (Charles C. Thomas, Springfield, IL, 1983).
- ⁴⁵F.M. Khan, *The Physics of Radiation Therapy* (Williams and Wilkins, Baltimore, MD, 1984).
- ⁴⁶J. van de Geijn, "Generalization of the projective high energy photon beam model," *Digest of the Fourth International Conference on Medical Physics; Physics in Canada* 32, p. 32.1 (1976).
- ⁴⁷J. van de Geijn, B. A. Fraass, R. W. Miller, and R. H. Creecy, "The projective beam model: an update on model and characteristic data," in *Proceedings of the 8th International Conference on the Use of Computers in Radiation Therapy*, pp. 94-98.1 (IEEE Computer Society, 1984).
- ⁴⁸J. A. Purdy, "Clinical photon beam dosimetry," in *Advances in Radiation Oncology Physics*, edited by J. A. Purdy (American Institute of Physics, New York, 1991).
- ⁴⁹E. McCullough, Chairman, Task Group 49, *Meterset Calculations, Radiation Therapy Committee* (American Association of Physicists in Medicine, College Park, MD, in progress).
- ⁵⁰C. G. Soares, "Fricke dosimetry in high-energy electron beams," NBS Special Publication 250-4 (U.S. Department of Commerce, National Institute of Standards and Technology, Washington, DC, 1987).
- ⁵¹F. H. Attix, "A protocol for the calibration of plane-parallel ion chambers by accredited dosimetry calibration laboratories," *Med. Phys.* 17, 931-933 (1990).
- ⁵²P. R. Almond, Chairman, Task Group 39, *Calibration of Plane-Parallel Chambers, Radiation Therapy Committee* (American Association of Physicists in Medicine, College Park, MD), *Med. Phys.* 21 (1994).
- ⁵³P. R. Almond, Chairman, Task Group 51, *Dosimetry Protocol for High Energy Photon and Electron Beams, Radiation Therapy Committee* (American Association of Physicists in Medicine, College Park, MD, in progress).
- ⁵⁴J. A. Purdy, Chairman, Task Group 46, *Accelerator Data, Radiation Therapy Committee* (American Association of Physicists in Medicine, College Park, MD, in progress).
- ⁵⁵D. Miller, Chairman, Task Group 23, *Verification of the Accuracy of Computerized Dose Calculation Systems for Photon Beams, Radiation Therapy Committee* (American Association of Physicists in Medicine, College Park, MD, in progress).
- ⁵⁶D. Meilenberg, Chairman, Task Group 52, *Asymmetric Collimators, Ra-*

- diation Therapy Committee (American Association of Physicists in Medicine, College Park, MD, in progress).
- ⁷J. R. Palta, I. Daftari, and N. Suntharalingam, "Field size dependence of wedge factors," *Med. Phys.* 15, 624-626 (1988).
- ⁸E. C. McCullough, J. Gortney, and C. R. Blackwell, "A depth dependence of the wedge transmission factor for 4-10 MV photon beams," *Med. Phys.* 15, 621-623 (1988).
- ⁹P. K. Kijewski, L. M. Chin, and B. E. Bjarngard, "Wedge-shaped dose distributions by computer-controlled collimator motion," *Med. Phys.* 5, 426-429 (1978).
- ¹⁰B. E. Bjarngard, "Computer controlled radiotherapy," in *Proceedings of the 1983 Mevatron Users Conference, Pebble Beach, CA* (Siemens Medical Systems, Iselin, NJ, 1983), pp. 177-209.
- ¹¹M. B. Levene, P. K. Kijewski, L. M. Chin, B. E. Bjarngard, and S. Hellman, "Computer controlled radiation therapy," *Radiology* 129, 769-775 (1978).
- ¹²D. D. Leavitt, M. Martin, J. H. Moeller, and W. L. Lee, "Dynamic wedge field techniques through computer-controlled collimator motion and dose delivery," *Med. Phys.* 17, 87-91 (1990).
- ¹³R. W. Miller, A. A. Raubitschek, J. van de Geijn, F. Orr, H. C. Xie, and E. Glatstein, "Treatment of mediastinal disease with mantle fields: Technical aspects and dosimetric considerations," *Int. J. Rad. Oncol. Biol. Phys.* 19, 192, abstract (1990).
- ¹⁴D. P. Fontenla, G. J. Kutcher, and T. J. Losasso, "Simulating blocks in treatment planning calculations," *Int. J. Rad. Oncol. Biol. Phys.* 16, 867-873 (1989).
- ¹⁵J. A. Meli, "Output factors and dose calculations for blocked x-ray fields," *Med. Phys.* 13, 405-408 (1986).
- ¹⁶American Association of Physicists in Medicine (AAPM), Task Group 25, "Clinical electron beam dosimetry: Report of the AAPM radiation therapy task group no. 25," *Med. Phys.* 18, 73-109 (1991).
- ¹⁷J. van de Geijn, B. Chin, J. Pockobradsky, and R. W. Miller, "A new model for computerized clinical electron beam dosimetry," *Med. Phys.* 14, 577-584 (1987).
- ¹⁸B. R. Thomadsen, L. W. Asp, J. van de Geijn, B. R. Paliwal, and C. PoCheng, "Perturbation of electron beam doses as a function of SSD due to the use of shielding blocks on the Clinac-18," *Med. Phys.* 8, 507-509 (1981).
- ¹⁹A. Niroomand-Rad, M. T. Gillin, R. W. Kline, and D. F. Grimm, "Dosimetry of small electron beams for routine radiotherapy planning," *Med. Phys.* 13, 416-421 (1986).
- ²⁰A. Niroomand-Rad, "Dosimetry of small elongated electron beams for treatment planning," *Med. Phys.* 16, 655-662 (1989).
- ²¹P. J. Biggs, A. L. Boyer, and K. P. Dopke, "Electron dosimetry of irregular fields on the Clinac 18," *Int. J. Rad. Oncol. Biol. Phys.* 5, 433-440 (1979).
- ²²P. J. Biggs, "The effect of beam angulation on central axis depth dose for 4-29 MeV; electrons," *Phys. Med. Biol.* 29, 1089-1096 (1984).
- ²³F. M. Khan, F. C. Deibel, and A. Soleimani-Meigooni, "Obliquity correction for electron beams," *Med. Phys.* 12, 749-753 (1985).
- ²⁴A. Jamshidi, F. T. Kuchnir, and C. S. Reft, "Determination of the source position for the electron beams for a high-energy linear accelerator," *Med. Phys.* 13, 942-948 (1986).
- ²⁵American Association of Physicists in Medicine (AAPM), *The Physical Aspects of Total and Half Body Photon Irradiation*, AAPM Report No. 17 (American Institute of Physics, New York, 1986).
- ²⁶J. Galvin, "Physical considerations for total body irradiation," in *Advances in Radiation Oncology Physics*, edited by J. A. Purdy (American Institute of Physics, New York, 1991).
- ²⁷E. W. Webster, "Physical considerations in the design of facilities for the uniform whole-body irradiation of man," *Radiology* 75, 19-32 (1960).
- ²⁸J. R. Cunningham, M. Woo, D. W. O. Rogers, and A. F. Bielajew, "The dependence of mass energy absorption coefficient ratios on beam size in phantoms," *Med. Phys.* 13, 496-502 (1986).
- ²⁹E. B. Podgorsak, C. Pla, M. D. C. Evans, and M. Pla, "The influence of phantom size on output, peak scatter factor and percentage depth dose in large field photon irradiation," *Med. Phys.* 12, 639-645 (1985).
- ³⁰J. VanDyk, P. M. K. Leung, and J. R. Cunningham, "Dosimetric considerations of very large cobalt-60 fields," *Int. J. Rad. Oncol. Biol. Phys.* 6, 753-759 (1980).
- ³¹J. VanDyk, "Broad beam attenuation of cobalt-60 gamma rays and 6, 18 and 25 MV x rays by lead," *Med. Phys.* 13, 105-110 (1986).
- ³²American Association of Physicists in Medicine (AAPM), *Total Skin Electron Therapy: Technique and Dosimetry*, AAPM Report No. 23 (American Institute of Physics, New York, 1987).
- ³³F. Khan, "Total skin electron therapy: Technique and dosimetry," in *Advances in Radiation Oncology Physics*, edited by J. A. Purdy (American Institute of Physics, New York, 1991).
- ³⁴A. Niroomand-Rad, M. T. Gillin, R. Komaki, R. W. Kline, and D. F. Grimm, "Dose distribution in total skin electron beam irradiation using the six-field technique," *Med. Phys.* 12, 415-419 (1986).
- ³⁵D. D. Leavitt, L. M. Peacock, F. A. Gibbs, and J. R. Stewart, "Electron arc therapy: Physical measurement and treatment planning techniques," *Int. J. Rad. Oncol. Biol. Phys.* 11, 987-999 (1985).
- ³⁶M. D. Mills, K. R. Hogstrom, and R. S. Fields, "Determination of electron beam output factors for 20 MeV linear accelerators," *Med. Phys.* 12, 473-476 (1985).
- ³⁷K. R. Hogstrom and D. D. Leavitt, "Dosimetry of arc electron therapy," in *Radiation Oncology Physics-1986*, edited by J. G. Kereiakes, H. R. Elson, and C. G. Born (American Institute of Physics, New York, 1986), p. 265-295.
- ³⁸F. M. Khan, G. D. Fullerton, J. M. F. Lee, C. V. Moore, and S. H. Leavitt, "Physical aspects of electron-beam arc therapy," *Radiology* 124, 497-500 (1977).
- ³⁹D. R. Rueggegger, S. D. Lerude, and D. Lyle, "Electron-beam arc therapy using a high energy betatron," *Radiology* 133, 483-489 (1979).
- ⁴⁰L. K. McNeely, G. M. Jacobson, D. D. Leavitt, and J. R. Stewart, "Electron arc therapy: Chest wall irradiation of breast cancer patients," *Int. J. Rad. Oncol. Biol. Phys.* 14, 1287-1294 (1987).
- ⁴¹D. D. Leavitt, L. Earley, and J. R. Stewart, "Design and production of customized field shaping devices for electron arc therapy," *Med. Dosimetry* 15, 25-32 (1990).
- ⁴²D. R. Rueggegger and D. Lyle, "Electron arc therapy using a high energy betatron," in *Proceedings of the Symposium on Electron Dosimetry and Arc Therapy*, edited by B. Paliwal (American Institute of Physics, New York, 1981), p. 223-248.
- ⁴³D. D. Leavitt, J. R. Stewart, J. H. Moeller, and L. Early, "Optimization of electron arc therapy doses by multivane collimator control," *Int. J. Rad. Oncol. Biol. Phys.* 11, 489-496 (1989).
- ⁴⁴M. Pla, C. Pla, and E. B. Podgorsak, "The influence of beam parameters on percentage depth dose in electron arc therapy," *Med. Phys.* 15, 49-55 (1988).
- ⁴⁵K. R. Kase and B. E. Bjarngard, "Bremsstrahlung dose to patients in rotational electron therapy," *Radiology* 133, 531-532 (1979).
- ⁴⁶F. M. Khan, "Calibration and treatment planning of electron beam arc therapy," in *Proceedings of the Symposium on Electron Dosimetry and Arc Therapy*, edited by B. Paliwal (American Institute of Physics, New York, 1982), p. 249-266.
- ⁴⁷B. E. Blackburn and R. Moreland, "An improved method of electron arc therapy for chest wall irradiation," *Int. J. Rad. Oncol. Biol. Phys.* 6, 1373, abstract (1980).
- ⁴⁸B. E. Blackburn, "A practical system for electron arc therapy," in *Proceedings of the Symposium on Electron Dosimetry and Arc Therapy*, edited by B. Paliwal (American Institute of Physics, New York, 1982), p. 295-314.
- ⁴⁹R. G. Kurup, K. R. Hogstrom, and M. E. Moyers, "Electron arc therapy: Comparison of pencil beam calculations with measurements," *Med. Phys.* 12, 509, abstract (1984).
- ⁵⁰A. L. Boyer, G. D., Fullerton, and J. G. Mira, "An electron beam pseudoarc technique for irradiation of large areas of chest wall and other curved surfaces," *Int. J. Rad. Oncol. Biol. Phys.* 8, 1969-1974 (1982).
- ⁵¹D. D. Leavitt, "Electron beam arc therapy," in *Advances in Radiation Oncology Physics*, edited by J. A. Purdy (American Institute of Physics, New York, 1991).
- ⁵²D. D. Leavitt, Chairman, Task Group 37, *Electron Arc Therapy*, Radiation Therapy Committee (American Association of Physicists in Medicine, College Park, MD, in progress).
- ⁵³B. A. Fraass, R. W. Miller, T. J. Kinsella, W. E. Sindelar, F. S. Harrington, K. Yeakel, J. Van de Geijn, and E. Glatstein, "Intraoperative radiation therapy at the National Cancer Institute: technical innovations and dosimetry," *Int. J. Rad. Oncol. Biol. Phys.* 11, 1299-1311 (1985).
- ⁵⁴J. P. Palta and N. Suntharalingam, "A non-docking intraoperative electron beam applicator system," *Int. J. Rad. Oncol. Biol. Phys.* 17, 411-417 (1989).
- ⁵⁵R. W. Piontek and K. R. Kase, "Design and dosimetric properties of an intraoperative radiation therapy system using an orthovoltage x-ray unit," *Int. J. Rad. Oncol. Biol. Phys.* 10, 1957-1965 (1984).
- ⁵⁶E. C. McCullough and J. A. Anderson, "The dosimetric properties of an applicator system for intraoperative electron beam therapy utilizing a Clinac-18 accelerator," *Med. Phys.* 9, 261-268 (1982).

- ¹⁰⁷P. J. Biggs, E. R. Epp, C. C. Ling, D. H. Novack, and H. B. Michaels, "Dosimetry, field shaping and other considerations for intra-operative electron therapy," *Int. J. Rad. Oncol. Biol. Phys.* 7, 875-884 (1981).
- ¹⁰⁸P. J. Biggs, W. U. Shipley, and C. Willett, "A method for shaping arbitrary fields for intraoperative radiotherapy," *Med. Phys.* 15, 445 (1988).
- ¹⁰⁹J. O. Archambeau, D. Aitken, T. M. Potts, and J. M. Slater, "Cost-effective, available-on-demand intraoperative radiation therapy," *Int. J. Rad. Oncol. Biol. Phys.* 15, 775-778 (1988).
- ¹¹⁰C. E. Nelson, R. Cook, and S. Rakfal, "The dosimetric properties of an intraoperative radiation therapy applicator system for a Mevatron 80," *Med. Phys.* 16, 794-799 (1989).
- ¹¹¹D. Jones, E. Taylor, J. Travagline, and S. Vermeulen, "A non-contacting intraoperative electron cone apparatus," *Int. J. Rad. Oncol. Biol. Phys.* 16, 1643-1647 (1989).
- ¹¹²K. R. Hogstrom, A. L. Boyer, S. M. Kirsner, and A. S. Shiu, "Design of metallic electron beam cones for an intraoperative therapy Linac," *Int. J. Rad. Oncol. Biol. Phys.* 18, 1223 (1990).
- ¹¹³E. C. McCullough and P. J. Biggs, "Intraoperative electron beam radiation therapy," in *Radiation Oncology Physics-1986*, edited by J. G. Kereiakas, C. G. Born, and H. R. Elson, AAPM Medical Physics Monograph No. 15 (American Institute of Physics, New York, 1987).
- ¹¹⁴E. McCullough, "Intraoperative Electron beam therapy (IORT)," in *Advances in Radiation Oncology Physics*, edited by J. A. Purdy (American Institute of Physics, New York, 1991).
- ¹¹⁵J. Palta, Chairman, Task Group 48, Intraoperative Electron Therapy, Radiation Therapy Committee (American Association of Physicists in Medicine, College Park, MD, in progress).
- ¹¹⁶R. A. Brown, T. S. Roberts, and A. G. Osbourne, "Stereotactic frame and computer software for CT-directed neurological localization," *Invest. Radiol.* 15, 308-312 (1980).
- ¹¹⁷L. Leksell, "Stereotactic radiosurgery," *J. Neurol. Neurosurg. Psychiatry*, 46, 797-803 (1983).
- ¹¹⁸J. T. Lyman, M. H. Phillips, K. A. Frankel, and J. I. Fabrikant, "Stereotactic frame for neuroradiology and charged particle Bragg peak radiosurgery of intracranial disorders," *Int. J. Rad. Oncol. Biol. Phys.* 16, 1615-1621 (1989).
- ¹¹⁹M. I. Hariz and A. T. Eriksson, "Reproducibility of the mounting of a non-invasive CT/MRI stereo-adapter," *Appl. Neurophysiol.* 49, 336-347 (1986).
- ¹²⁰S. S. Gill, Head Fixation Apparatus, U.K. Patent Application G. B. 2213066A (1989).
- ¹²¹C. B. Saw, K. Ayangar, and N. Suntharalingam, "Coordinate transformation and calculation of angular and depth parameters for a stereotactic system," *Med. Phys.* 14, 1042-1044 (1987).
- ¹²²R. L. Siddon and N. H. Barth, "Stereotaxic localization of intracranial targets," *Int. J. Rad. Oncol. Biol. Phys.* 13, 1241-1246 (1987).
- ¹²³L. Steiner, "Treatment of arteriovenous malformations," in *Intracranial Arteriovenous Malformations*, edited by C. B. Wilson and M. B. Stein (Williams and Wilkins, Baltimore/London, 1984), p. 295-313.
- ¹²⁴D. D. Leavitt, F. A. Gibbs, M. P. Heilbrun, J. M. Moeller, and G. A. Takach, Jr., "Dynamic field shaping to optimize stereotactic radiosurgery," *Int. J. Rad. Oncol. Biol. Phys.* 21, 1247-1255 (1991).
- ¹²⁵T. M. Peters, J. A. Clark, G. B. Pike, H. L. Collins, D. Leksell, and O. Jeppsson, "Stereotactic neurosurgery planning on a personal computer-based work station," *J. Digit. Imag.* 2, 75-81 (1989).
- ¹²⁶W. A. Friedman and F. J. Bova, "The University of Florida Radiosurgery System," *Surg. Neurol.* 32, 334-42 (1989).
- ¹²⁷W. A. Friedman and F. J. Bova, "Stereotactic radiosurgery," *Contemp. Neurosurg.* 11, 109 (1989).
- ¹²⁸G. Chierego, C. Marchetti, R. Avanzo, F. Pozza, and F. Colombo, "Dosimetric considerations on multiple arc stereotaxic radiotherapy," *Radiother. Oncol.* 12, 141-152 (1989).
- ¹²⁹L. Walton, C. K. Bomford, and D. Ramsden, "Sheffield stereotactic radiosurgery unit: physical characteristics and principles of operation," *Br. J. Radiol.* 60, 897-906 (1987).
- ¹³⁰P. V. Houdek, J. V. Fayos, J. M. Van Buren, and M. S. Ginsberg, "Stereotaxic radiotherapy technique for small intracranial lesions," *Med. Phys.* 12, 469-472 (1985).
- ¹³¹G. H. Hartmann, W. Schlegel, V. Sturm, B. Kober, O. Pasty, and W. J. Lorenz, "Cerebral radiation surgery using moving field irradiation at a linear accelerator facility," *Int. J. Rad. Oncol. Biol. Phys.* 11, 1185-1192 (1985).
- ¹³²F. Colombo, A. Bennedetti, F. Pozza, R. C. Avanzo, C. Marchetti, G. Chierego, and A. Zanando, "External stereotactic irradiation by linear accelerator," *Neurosurgery* 16, 154-160 (1985).
- ¹³³L. Lunsford, J. Flickinger, G. Lindner, and A. Maitz, "Stereotactic radiosurgery of the brain using the first United States ⁶⁰Co source gamma knife," *Neurosurgery* 24, 151-159 (1989).
- ¹³⁴E. B. Podgorsak, A. Oliver, M. Pla, P. Lefebvre, and J. Hazel, "Dynamic stereotactic radiosurgery," *Int. J. Rad. Oncol. Biol. Phys.* 14, 115-126 (1988).
- ¹³⁵P. V. Houdek, J. M. Van Buren, and J. V. Fayos, "Dosimetry of small radiation fields for 10-MV x-rays," *Med. Phys.* 10, 333-336 (1983).
- ¹³⁶R. J. Rice, J. L. Hansen, G. K. Svenson, and R. L. Siddon, "Measurements of dose distributions in small beams of 6 MV x-rays," *Phys. Med. Biol.* 32, 1087-1099 (1987).
- ¹³⁷O. Betti, C. Munari, and R. Rosier, "Stereotactic radiosurgery with the linear accelerator: treatment of arteriovenous malformations," *Neurosurgery* 24, 311-321 (1989).
- ¹³⁸W. Lutz, K. R. Winston, and N. Maleki, "A system for stereotactic radiosurgery with a linear accelerator," *Int. J. Rad. Oncol. Biol. Phys.* 14, 373-381 (1988).
- ¹³⁹E. B. Podgorsak, A. Olivier, M. Pla, P.-Y. Lefebvre, and J. Hazel, "Dynamic stereotactic radiosurgery," *Int. J. Rad. Oncol. Biol. Phys.* 14, 115-126 (1988).
- ¹⁴⁰B. Pike, E. B. Podgorsak, T. M. Peters, and C. Pla, "Dose distribution in dynamic stereotactic radiosurgery," *Med. Phys.* 14, 780-789 (1987).
- ¹⁴¹E. B. Podgorsak, G. B. Pike, A. Oliver, M. Pla, and L. Souhami, "Radiosurgery with high energy photon beams: A comparison among techniques," *Int. J. Rad. Oncol. Biol. Phys.* 16, 857-865 (1989).
- ¹⁴²F. J. Bova, "Stereotactic radiosurgery," in *Advances in Radiation Oncology Physics*, edited by J. A. Purdy (American Institute of Physics, New York, 1991).
- ¹⁴³M. Schell, Chairman, Task Group No. 42, Stereotactic Radiosurgery, Radiation Therapy Committee (American Association of Physicists in Medicine, College Park, MD, in progress).