Verification of monitor unit calculations for non-IMRT clinical radiotherapy: Report of AAPM Task Group 114

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The requirement of an independent verification of the monitor units (MU) or time calculated to deliver the prescribed dose to a patient has been a mainstay of radiation oncology quality assurance. The need for and value of such a verification was obvious when calculations were performed by hand using look-up tables, and the verification was achieved by a second person independently repeating the calculation. However, in a modern clinic using CT/MR/PET simulation, computerized 3D treatment planning, heterogeneity corrections, and complex calculation algorithms such as convolution/superposition and Monte Carlo, the purpose of and methodology for the MU verification have come into question. In addition, since the verification is often performed using a simpler geometrical model and calculation algorithm than the primary calculation, exact or almost exact agreement between the two can no longer be expected. Guidelines are needed to help the physicist set clinically reasonable action levels for agreement. This report addresses the following charges of the task group: (1) To re-evaluate the purpose and methods of the "independent second check" for monitor unit calculations for non-IMRT radiation treatment in light of the complexities of modern-day treatment planning. (2) To present recommendations on how to perform verification of monitor unit calculations in a modern clinic. (3) To provide recommendations on establishing action levels for agreement between primary calculations and verification, and to provide guidance in addressing discrepancies outside the action levels. These recommendations are to be used as guidelines only and shall not be interpreted as requirements. [DOI: 10.1118/1.3521473]

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I. INTRODUCTION

The delivery of therapeutic radiation is a medical procedure and as such requires independent confirmation to ensure correct and accurate delivery. This confirmation is accomplished by a comprehensive calculation and chart review procedure performed both before and throughout patient treatment.[1–4] A key component of this procedure, and the focus of this report, is the "independent monitor unit verification," where the monitor unit (MU) setting determines the quantity of radiation delivered to the patient. In this report, the "primary MU" refers to the calculation used for the actual treatment of the patient, while "MU verification" (MUV) refers to a measurement or calculation that is performed only for the purpose of confirming the primary MU calculation and is not used for the delivery of radiation. Consistent with other American Association of Physicists in Medicine Task Groups,[1,5] the MUV is considered to be an independent check of the dose or machine setting at one point and is a separate activity distinct from the treatment plan review, with its own assessment of the veracity of the factors that are used in the check. While these settings are generically referred to as "monitor units" in this report, this term also applies to time settings for treatment units that employ radioactive sources.

Radiation therapy has changed significantly over the past years. In the past, both primary and verification calculations were performed using manually derived and entered data, and the need for a check was obvious in order to identify transcription errors, depth misreadings, or the incorrect look-up of a table parameter. A second person independently verified the patient information such as source to surface distance (SSD) and depth, parameters such as TMR and output factor, and the calculation arithmetic. Agreement between the MUV and the primary MU calculations to within a set tolerance was easily achieved due to the application of almost identical methods for both calculations.

The introduction of extensive computerization, volumetric imaging, and improved computation algorithms has changed both the complexity of the patient treatment and the manner in which MUV is performed. Most MUV calculations are performed today by computer programs using electronic data transfer protocols, which are much less likely to result in the arithmetic, transcription, or look-up errors that the traditional verification calculation was designed to find. Modern treatment planning systems (TPSs), which use sophisticated algorithms and build 3D geometrical patient models complete with heterogeneous tissue densities, are complex. This complexity presents challenges to traditional manual verification methods since much of the information required to perform the verification needs to come directly from the TPS.

These changes have brought into the question the value of performing the MUV. There has been anecdotal speculation whether any further checks should be required for a planning system which has been carefully and thoroughly commissioned and passed a series of quality assurance (QA) tests. A physicist's time is a limited resource in a busy radiotherapy department, and performing a verification of every calculation for every patient can be time consuming and a major drain on this resource. It is critical to evaluate the procedures that are performed to ensure that they are not only necessary but also effective. These types of questions are best addressed by identifying both the value and the limitations of the MUV in the modern clinic; such an assessment by this task group is provided in Secs. II and III in the context of

recent treatment incident reports. The task group concludes that the MUV remains a useful and necessary step in assuring safe and accurate patient treatment and discusses aspects of an MUV program in Sec. IV.

The clinical physicist needs guidelines for acceptability of agreement between the primary and verification calculations. In the days of 2D and manual primary calculations, essentially perfect agreement was expected since a virtually identical methodology was repeated for the MUV calculation. Now, with the use of different algorithms that may incorporate different approximations for both the patient and the beam parameters, the acceptable level of agreement that can be expected is unclear. This is further complicated by the fact that the verification algorithm is usually simpler, and there is a tendency to attribute large discrepancies to the simplicity of the algorithm.[6] While larger discrepancies may be expected, procedures and guidelines should be established to eliminate, as much as practical, sources of discrepancies, and methodologies should be established to resolve the larger discrepancies. Section VI provides suggested guidelines for establishing agreement action levels, as well as remedial actions that can be undertaken if the difference between the primary MU calculation and the MUV exceeds these levels.

Given this environment, the present task group was formed to:

(1) Re-evaluate the purpose and methods of the "independent second check" for monitor unit calculations for non-intensity modulated radiation therapy (IMRT) treatment in light of the complexities of modern-day treatment planning.
(2) Present recommendations on how to perform MUVs in a modern clinic.
(3) Provide recommendations on establishing action levels for agreement between primary calculations and verification, and to provide guidance in addressing discrepancies outside the action levels. These recommendations are to be used as guidelines only and shall not be interpreted as requirements.

This report does not address or apply to the verification of IMRT calculations. The AAPM has created a separate working group on IMRT to consider IMRT verification and other IMRT-related issues.

II. OBJECTIVE OF THE VERIFICATION CALCULATION

The goal of the MUV is to ensure that the primary monitor unit calculation is sufficiently accurate for the safe and effective treatment of the patient. While the accuracy and safety of the primary calculation system, typically a TPS, is verified to a large extent by a thorough and complete set of acceptance and commissioning tests, it is recognized that most centers do not have the resources to perform a comprehensive set of tests on their own.[5] It was recognized by other task groups that commissioning tests, designed to establish the accuracy limits of the system, cannot test the system under all possible current and future clinical scenarios;[5] this is particularly evident given the increased functionality and complexity added to planning systems in the past few years. Typically commissioning tests only cover the existing practices at a particular center. Subtle changes in process could lead to the use of the untested modules within the TPS or usage in a manner that was not originally tested. These limitations necessitate the performance of a MUV to ensure appropriate dose delivery in each instance of a TPS calculation.

There are a number of specific incidents that point to the value of a MUV. For example, in 2006, a patient received a fatal radiation overdose due to the introduction of electronic data transfer between the electronic patient chart, the TPS, and the treatment delivery system. Electronic data transfer was adopted for most treatment sites, with the exception of some specialized and infrequently used techniques; a manual MU calculation based on a particular plan normalization style was used for the treatment even though the TPS generated plan used a different normalization. The investigation into this event indicated that an independent calculation check may have identified the resulting overdose prior to treatment.[7] In France, between 2004 and 2005, 23 patients received an overexposure of radiation resulting from the introduction of the enhanced dynamic wedge into the clinic;[8] a review of the incident pointed to the recent elimination of an independent check of the MU calculation as a major contributing factor. A similar deficiency in procedure was identified in an incident in Panama in 2000, where 28 patients were overexposed to radiation due to the incorrect usage of the TPS. An International Atomic Energy Agency (IAEA) review of the incident[9] included among its recommendations:

> *"Results provided by the TPS need to be checked, and this should include verification by manual calculation of the treatment time and dose to the selected point. This verification should be part of the QA programme."*

The IAEA, in a review of radiotherapy mistreatments[10] prior to these events, concluded, in part, that an independent verification of the treatment unit settings was an important element required to ensure patient safety. More recent reviews of radiotherapy treatment errors support this assessment through an acknowledgment that the complexity of modern TPSs leads to a larger opportunity for mistreatment and misuse, which is best guarded against through a set of rigorous QA procedures. The ICRP, in a 2009 report aimed at preventing exposure errors from new treatment technologies, states *"A simple secondary MU calculation, independent from the TPS, has proven for many years to be an efficient tool for prevention of major errors in dose delivery."*[11] These reviews of incidents all point out that checks such as MUV are important for patient safety.

An effective MUV is one of the several tools in the QA process designed to catch gross errors in the treatment dose delivery for an individual patient. An evaluation of the inci-

dence of radiotherapy errors over 10 years at a large regional cancer center concluded that treatment plan checks, including MU verification calculations, were very effective in detecting documentation and treatment planning errors.[12] The types of errors most likely to occur in a particular center will vary with staffing and process details. The responsible physicist at each center is the person best suited to assess the type of errors that can occur during the planning and delivery process. The physicist should ensure that the MU verification process effectively detects the most probable causes of gross errors and utilizes reasonable action levels for disagreement between primary calculations and the verification check. This responsibility extends to maintaining clear documentation and providing periodic training for staff in all disciplines that are required to review and understand the MUV, including both the limitations of the calculation and the established action levels for this QA activity.

II.A. Types of error

A verification calculation necessarily will generate results that vary from the primary calculation. Small differences on the order of a few percent do not necessarily indicate an error in the primary calculation. Such discrepancies may result from differences in algorithms, in geometrical patient representation, or in small variations in beam data. Larger differences, however, may be a symptom of an error. While there are too many sources of potential error in monitor unit calculations to compile a comprehensive list, it is useful to classify errors into two broad types, here referred to as random and systematic errors. This classification will help in identifying when these errors occur and minimizing the chances for their occurrence.

A random error is not reproducible, by its very definition. If the calculation is repeated, especially by a different person, the error is likely to be detected. Examples of random errors are incorrect beam energy, wrong dose, manual arithmetic calculation errors, incorrect dosimetric data retrieval from a table, incorrect bolus thickness characterization, and incorrect manual transcription of field size. Random errors can effectively be reduced by a MUV calculation. The chance of the same random error repeating itself in both the primary and the verification calculation is, by definition, extremely small.

The second category of error is due to systematic causes, generally attributed to a defect in part of the calculation procedure.[13] There is a correlation between the circumstance in which it occurs and its chance of occurrence. The error will likely occur again for the same calculation even if performed by a different person. One class of systematic errors results from misunderstanding of standard procedures; for example, the misinterpretation of plan dose normalization or beam calibration designation (e.g., depth of d_m vs 10 cm, SSD vs SAD). Other examples of systematic errors are an incorrect entry in the tissue phantom ratio (TPR) table used for calculation, an error in the standard form for a manual calculation, and an algorithmic error in, or misapplication of, the software used in monitor unit calculations. The effectiveness of identifying systematic errors through a verification check that is comprised of repeating the same calculation by an independent person is more limited than for random errors; rather, the use of a different calculation program/algorithm or a measurement is more likely to find such an error. Systematic errors are best reduced by a thorough review of the process, a careful commissioning procedure, and complete and accurate data tables. An external audit of the clinic's process, such as performed by the Radiologic Physics Center for institutions involved in NCI-funded protocols or offered by M.D. Anderson's Radiation Dosimetry Services for beam calibration, is recommended as a further method to reduce systematic errors.

An error that occurs in the course of a MU calculation might be either random or systematic, depending on the process. For example, the SSD can be erroneously reported and used by a planning system due to some thermoplastic mask material stretching across an air space. If the standard procedure of the institution includes a step to use tools within the planning system to check the automatically detected SSD and this step was missed, then a random error has occurred. However, if the standard procedure does not include such a check before the MU calculation, this would be a systematic error in the treatment planning process.

The increasing use of computer programs for MU calculations will shift certain types of errors from random to systematic. It is highly unlikely that after appropriate commissioning, a computer program will incorrectly look-up data at random, download a wrong field size intermittently, and make other similar errors. This does not mean that computerized MU calculations will generate error-free results; virtually all programs contain coding errors and are built around assumptions of usage that can potentially lead to an incorrect result. The pattern and procedures that give rise to anomalous results can provide information that can isolate the underlying cause of the problem, which could be a code logic error, a data or database error, or inappropriate program use.

II.B. Potential errors

The level of radiotherapy computer systems connectivity present in an institution plays a critical role in the frequency and type of error that may occur. This interplay of connectivity and error frequency is illustrated in Table I, which lists some common errors encountered in the past and rates their occurrence probability according to connectivity. The connectivity among CT-simulator, TPS, treatment chart, record–and-verify system and treatment delivery systems (including block cutters) are classified into three different groups:

(1) Manual—no connectivity, all parameters for treatment must be manually copied into a treatment chart with a relatively high possibility of transcription errors.
(2) Partial connectivity—connectivity between some systems, or only basic parameters are transferred, but some

TABLE I. Likelihood of potential errors in MU calculation.

Potential error condition	Degree of automation			Comment
	Manual	Partial	Full	
(1) Tray factors				
(a) Leaving out tray	H[a]	M[a]	L[a]	
(b) Using wrong tray	H	H	L	
(2) Wedge factors				
(a) Leaving out wedge	H	M	L	
(b) Wrong wedge	H	L	L	
(c) Wrong wedge direction (for enhanced dynamic wedge or for off-axis calculations)	H	L	L	
(d) Off-axis in wrong direction (e.g., along instead of perpendicular to wedge gradient)	H	L	L	
(3) Not planning according to Rx				
(a) Wrong energy	H	L	L	
(b) Wrong field size	H	M	L	
(c) Wrong beam weights	H	L	L	
(d) Wrong prescription dose	H	L	L	
(4) Wrong depth	M	M	M	
(5) Wrong equivalent square for open or blocked field size	M	L	L	
(6) Wrong SSD	M	M	M	
(7) Calculation point too close to a field or block edge (e.g., in penumbra)	M	M	M	
(8) Wrong CT data set	L	H	H	Assumed that conventional simulation will be used in manual, resulting in check.
(9) Incorrect automatic contouring of CT images, including replacing sim table with treatment table	L	M	M	
(10) Wrong density derivation from CT images	M	M	M	Includes effects from radiographic contrast material, extrapolation of conversion table
(11) Wrong scaling of CT/MR images	L	M	M	Issue more related to imaging system QA
(12) Inconsistencies/errors in input data	M	M	M	
(13) Errors arising from incorrect documentation of treatment (e.g., bolus not documented)	M	M	L	
(14) Errors arising from not changing defaults to actual patient settings in treatment planning system (e.g., energy, field size, machine)	M	M	L	Assumes connectivity would catch such errors
(15) Computer bugs	M	H	H	Complexity of systems increases chance of bug
(16) Data transmission errors	M	M	L	Assumes separate data sources for manual and partial automation categories, increasing chance of inconsistency
(17) Data corruption	M	L	L	Transcription errors in first case
(18) Change to already approved plan	M	M	M	
(19) Using untested area of treatment calculation system	L	H	H	
(20) Calculating with wrong grid size (too coarse, might be especially a problem for electrons)	M	M	M	Requires tight tolerances to catch

[a]H = high, M = medium, and L = low likelihood of potential error.

manual data entries are required. Examples of required adjustments include prescription, set MU, wedge orientation, MLC shapes, and tray types and positions.

(3) Full integration—complete transfer of all information including custom shaping and wedge orientation. Adjustment of field parameters occurs only for clinically based changes requested after the treatment has been planned.

The majority of centers likely fall into category (2), with many centers approaching category (3). In these environments, the probability of an event can be classified as low (L), medium (M), or high (H). Table I is a task group member consensus analysis of typical events and is intended to be used as an initial guide for centers performing their own process review, rather than as a comprehensive list of all

possible errors and their likelihood of occurrence. The likelihood of encountering any particular error will strongly depend on the details of the processes implemented at any specific center.

III. LIMITATIONS OF THE VERIFICATION CALCULATION

The verification calculation is not and should not be used as a check of the overall accuracy of the primary TPS; that is the function of commissioning and continual QA. It is crucial that both the primary and the verification planning systems be properly and thoroughly commissioned so that they are as accurate as possible. Monitoring the agreement between the TPS and the verification system during clinical use can aid in identifying regions where beam models or data may be improved, but such monitoring is not a substitute for the commissioning of either system. Both the TPS and the verification system should be fully tested and commissioned following accepted guidelines[5,14,15] prior to clinical use.

A key result from commissioning tests is the establishment of the expected accuracy of the primary TPS and MUV system in clinically relevant situations. This provides the basis for the action levels that are set for the MU verification calculation. TPS accuracy will be clinic-dependent and is expected to vary with the complexity of the calculations. The verification program is typically not as accurate as the primary TPS, particularly for complex geometries and/or heterogeneity corrections. Therefore, the agreement of the verification and primary calculations should be expected to vary with the complexity of the situation. See Sec. VI for further discussion of the expected agreement and action levels.

The MUV is not a check of the accuracy of the entire calculated dose distribution. It only verifies that the monitor units determined by the primary TPS will deliver the expected dose within acceptable uncertainty to a single point within the treatment volume. That point may be the isocenter, the prescription point, or some other well-defined points (see Sec. V A on the selection of the comparison point). The commissioning and periodic check of the TPS calculation algorithm must be relied upon to assure that the accuracy at the comparison point implies accuracy throughout the targeted tissues.

Since the verification is confined to checking the dose calculation at a single point in the field, the verification by itself does not constitute a complete QA plan review. The MUV is only one part of a complete physicist's plan review. While many elements of a plan review are implicit in the MUV, a careful review of all the plan parameters common to the MUV must still be performed to ensure these data reflect the actual patient treatment. Thus, in addition to the MUV, the physicist's plan review should confirm that the dose, beam energy, fractionation, and dose point location are consistent with the physician's prescription. Other data such as effective equivalent field can be estimated from the exposed area seen in a DRR or MLC pattern. These and other elements common to both the physicist's plan review and the verification data, such as beam SSD, beam modifier identification, and orientation, need to be examined in the context of the intended treatment to ensure that a systematic error in the plan is not propagated to the MUV. In the event that the difference between the MUV and primary calculation is outside the accepted action levels, the resolution of the discrepancy should include a re-examination of the parameters used in the calculation that are directly derived from the TPS.

IV. ASPECTS OF A MU VERIFICATION PROGRAM

In most institutions, MU verification is performed using a computer program or TPS different from the primary TPS, although manual calculations may occasionally be used. The verification program is usually, but not necessarily, less complex than the primary TPS, using simpler calculation algorithm and patient geometrical representation.

IV.A. MU verification and patient plan quality assurance

The MUV is one component of a complete plan review performed by a qualified medical physicist to ensure the safety and efficacy of a treatment plan. The purpose of the plan review is to ensure that the treatment plan is clinically reasonable, that dosimetric calculations are correct, and that the plan will deliver treatment as specified in the prescription. Guidelines for plan checks are given in AAPM Task Group 40 (Ref. 1) and by the American College of Radiology.[3] This in-depth review should consider all elements of the treatment plan. The reviewer must verify that correct parameters are used in MU calculations, and that calculations are performed correctly as per the physician's prescription. The reviewer must also confirm that treatment setup and delivery information are transcribed properly from physician's intent through simulation to treatment planning and from treatment planning to the patient's chart and/or the record-and-verify system.

This task group recommends that the verification calculation should be completed prior to the delivery of the first treatment fraction. In the rare occasion where this is not possible, the MUV should be performed as soon as possible, consistent with the recommendations of TG-40: Prior to the delivery of the third treatment fraction or 10% of the prescribed dose, whichever corresponds to the least delivered dose.

Each center performing after-hours emergency treatments should have a policy detailing how planning, including the MUV, is to proceed when a physicist is not immediately available for plan and calculation review. The staff designated to calculate the treatment MU, usually the radiation therapist or radiation oncologist, must be adequately trained in the use of calculation software and/or dosimetry data tables, and that data must be readily accessible. The MUV should then be performed as soon as possible, preferably before the next treatment but in any event conforming to the recommendations of TG-40 stated above. This task group also recommends the policy includes a maximum fractional dose that may be delivered in emergency treatments prior to

a complete physics review and MUV, and suggests a value of 3–4 Gy. Finally, the policy should include periodic training to maintain the competency of staff involved in after-hours treatments.

MUV is an important part of the clinical physicist's responsibilities, and adequate time must be allotted for this task. Hospital and departmental administrators must provide adequate physics staffing to allow for the timely completion of patient plan QA checks, including the MUV, in conformance with the TG-40 recommendations. Administrators are referred to the ACR/AAPM Abt Study of Medical Physics Work Values for Radiation Oncology Services[16] for guidelines on these activities.

IV.B. Commissioning of a verification system

As with any system used in the clinical treatment of patients, the MUV system requires commissioning and ongoing quality control monitoring to ensure the accuracy, safety, and efficacy of the system. While the details of the commissioning process for a clinical dosimetric system are outside the scope of this report, it is important to note that the MUV system constitutes such a system and requires its own comprehensive commissioning.

For the purposes of commissioning, the MUV system should undergo a testing procedure for conventional TPSs such as has been recommended by others.[5,14,15] Although these reports focus on 3D TPSs, which typically have more complex functionality than MUV systems, the core dosimetric tests described are applicable to the commissioning of the MUV system. The calculated dose resulting from a given number of monitor units should be independently verified by measurements and compared, if possible, with the results of other established calculation systems of known accuracy, such as institutional data previously validated by independent measurement, RPC site visit, or other means. Commissioning tests for the system should include clinically relevant geometries that verify the accuracy of shaped field calculations and calculations in heterogeneous media if such situations are encountered in the clinical practice of the institution. This testing program should establish the accuracy of the MUV system in different clinical situations, and this determination should be used to establish action levels as described in Sec. VI A of this report. Once commissioned, a QA program should be implemented for continual surveillance of the constancy of the system. Additional checks may be done to ensure that the mathematical models and data used for a given treatment machine model and beam energy closely match available published data.

Other nondosimetric clinical functionalities within a MUV system require commissioning as well. As with a conventional TPS, the configuration and presentations of the treatment unit geometry within a MUV system should undergo specific testing. In particular, the crucial functionality of plan data import and transfer needs to be carefully and thoroughly tested to ensure the integrity and completeness of the data passed between systems if it is to be used as part of the clinical process.

IV.C. Independence of verification calculation

An "independent" check of the dose calculation is called for by several professional groups.[1,3] The meaning of "independence," however, has to be carefully considered in the modern environment of sophisticated computer calculation algorithms and direct communication between TPS, MUV calculation systems, and treatment delivery systems.

In general, the independence of the MUV is established by using a different methodology and/or program than that used for the primary calculation.[4] The only exception to this principle applies to a manually calculated verification of a primary manual calculation. An independent calculation cannot be obtained by using the same program for both the primary and the verification calculations. Even if a TPS has more than one calculation model implemented, the use of a separate program is strongly recommended over the use of a different beam model within the same system because many of the potentially errant parameters would be common to both calculation models. The verification program typically uses a different beam and/or patient model than the primary TPS. In the case when the beam and patient models are similar, the algorithmic implementations should be different. This is typical if the programs were developed by different vendors.

The files containing beam data and parameters used by the verification calculation program should be separate and independent of the files used by the primary calculation,[4] even if both sets of files are based on the same measured data. This will aid in the identification of calculation errors due to data errors (either incorrect/inaccurate data or typographical mistakes) as well as errors due to corruption of the stored values.

The electronic transfer of patient treatment parameters from the primary TPS to the verification program, either directly or via the record-and-verify system, is encouraged and does not invalidate the independence of the verification calculation. The chance of electronic transfer errors during download is much less than the chance of manual entry transcription errors. However, a careful examination of all the transferred parameters is crucial to ensure that the data reflect an accurate description of the physical treatment conditions and do not replicate an error such as an incorrectly determined SSD/depth or dose artifact from poor calculation point location. Whenever possible, parameters should be independently verified or determined from basic planning data, such as extracting the field size and blocked equivalent square from the beam's eye view (BEV) and obtaining the treatment depth independently of the CT image information. For some parameters, such as radiological depth, only a rough estimation may be possible. The effectiveness and independence of the MUV relies on the assurance that electronically transferred calculation parameters are independently verified to accurately describe the treatment conditions.

The MU verification calculation should be performed preferably by an independent physicist or under physics supervision by a qualified individual who is not involved in the

TABLE II. Guidelines for action levels for disagreement between verification and primary calculations for homogeneous conditions.

	Similar calculation algorithms			Different calculation algorithms		
Primary calculation geometry	Same patient geometry (%)	Approx. patient geometry (%)	Uniform cube phantom approx. (%)	Same patient geometry (%)	Approx. patient geometry (%)	Uniform cube phantom approx. (%)
Minimal field shaping	2	2.5	3	2.5	3	3
Substantial field shaping and/or contour change	2.5	3	4	3	3.5	4
Wedged fields, off-axis	2	2.5	3	3.5	4	5

primary calculation. In some practices, it is the treatment planner who electronically transfers parameters from the primary TPS to the verification system and possibly even performs the verification calculation itself. In such cases, a careful review of both calculations must be done by an independent physicist or under physics supervision by a qualified individual who is not involved in the primary calculation.[17]

IV.D. Potential advantages of a measurement-based verification algorithm

While any valid dosimetric calculation system, up to and including a second TPS or Monte Carlo simulation, can be used to perform a verification MU calculation, there is a potential benefit to using a simple measurement-based calculation system in which the effects of scatter, missing tissue, and tissue heterogeneity are separated and can be independently assessed. While a state-of-the-art TPS may provide a very good simulation of the dose distribution, it must be recognized that it is nonetheless a simulation using derived parameters that can be quite separate from the original input dosimetry measurements and can behave in unexpected ways. A measurement-based verification algorithm balances the complexity of the primary TPS with the transparency of a calculation based directly on a set of simple measurements, with a few well understood small perturbations applied to account for patient geometry effects.

A system comprised of measured parameters, such as is described in Appendix A, enables the physicist to decompose a calculation and consider the impact of each factor on an individual basis. Such an approach enables the physicist to isolate the potential causes of a discrepancy, relate any investigative measurements to calculations, and thus aid in identifying the underlying cause of the discrepancy. The approach may be limited, however, for very complex calculations involving heterogeneity corrections.

V. RECOMMENDED METHODS OF MU VERIFICATION

The parameters and formalism used in the MUV depend on several factors such as radiation type, beam energy, algorithms used by the primary and verification calculation systems, irregularity of the treatment field, location of the comparison point within that field, the field-shaping device (blocks, MLC), use of different beam modifiers (wedges, compensators, etc.), and the tissue heterogeneities. The treatment machine normalization point plays a key role in the MUV formalism. In addition, the attainable verification accuracy is critically dependent on the selection of the comparison point within the field and patient geometry. Note that while either an in-phantom dose measurement or an independent calculation of the dose for a given MU or the MU required to deliver a particular dose can serve as a valid MUV, the following discussion focuses primarily on the use of a calculation methodology.

V.A. Comparison point selection

The appropriate placement and selection of a comparison point is critical in achieving an accurate verification and, consequently, achieving an agreement within the anticipated limits listed in Tables II and III. It is recommended that placement of the plan normalization point should follow ICRU guidelines[18,19] for dose reporting. Although preferable, the MU verification point need not coincide with the plan normalization point. It is sufficient, for verification purposes,

TABLE III. Guidelines for action levels for disagreement between verification and primary calculations with heterogeneity corrections.

	Similar calculation algorithms		Different calculation algorithms	
Primary calculation geometry	Same patient geometry (%)	Approx. patient geometry (%)	Same patient geometry (%)	Approx. patient geometry (%)
Large field	2	3	2.5	3.5
Wedged fields, off-axis	2	3	3.5	4.5
Small field and/or low-density heterogeneity	3	3.5	4	5

to select an alternate point which may provide more straightforward computation and still demonstrate agreement between the primary and MU verification methodology. Additionally, although it is desirable to use a single point within the patient for comparison in all fields, this may not be feasible in many circumstances, as a selected point in the patient may be shielded from some beams. In such situations, separate points may be required for different fields.

The dosimetry in regions of electronic disequilibrium carries additional uncertainties and should be avoided during point placement. This implies that point placement in regions containing a high dose gradient should be avoided; as well, a calculation point should not be placed near a field edge. Conservatively, points within 2 cm of a field edge may experience disequilibrium effects arising from lack of lateral scatter. Patient geometry may also give rise to regions of electronic disequilibrium when calculations incorporate corrections for heterogeneous tissue densities.[20,21] Since regions of electronic disequilibrium are found near tissue interfaces, calculation points should be located, if possible, in soft tissue and positioned at least 1.0 cm downstream and 1.0 cm lateral to tissue (heterogeneous media) interfaces to avoid large disequilibrium effects. The ray-line path from the source to the calculation point should provide a representative sample of the bulk properties experienced by the treatment field; nonrepresentative ray-line paths such as paths through small heterogeneities like the trachea or passing tangentially through the edge of a long bone disproportionately accentuate the average density change along the path length and should be avoided.

V.B. Differences in primary verification and calculation methods

The primary MU calculation is typically based on a detailed model of the fields and patient geometry. The primary calculation from a modern 3D TPS accurately accounts for patient scatter effects arising from blocked fields, scatter changes arising from contour effects, and patient heterogeneity effects. The verification MU calculation in general will not have as detailed a field or patient model as the primary MU calculation and, consequently, will give rise to the necessity for an acceptance range between these two calculations. The factors that are likely to introduce the greatest uncertainty in the verification calculation are blocked field scatter, patient contour, and patient heterogeneity effects.

Historically, many verification calculations did not take into account the effects of contour and heterogeneity, or used planning system derived values for these corrections. For the purposes of this discussion, this type of calculation is referred to here as a "uniform cube phantom" geometry. In other cases, an approximate model of patient geometry was applied in the form of simple correction factors such as a scatter correction for tangentially incident fields or path length type heterogeneity corrections; such calculations are referred to as "approximate patient" geometry. If both the primary MU and the validation calculation methods use the same methodologies and approximations for obtaining the MU setting, then no additional uncertainties should arise. An example of such a situation is the use of a manual look-up table calculation for both the primary and the MUV calculations; another example is when both the primary and the verification calculations use programs that have implemented a Clarkson type calculation in the uniform cube phantom geometry. Both situations will be referred to in this discussion as having the "same patient geometry."

V.C. Manual verification methods

Manual calculational methods using dosimetry look-up tables are very mature and provide the basis for most MUV calculations, either through a manual process or as input into computer programs. All radiation therapy clinics should maintain an accurate set of dosimetry tables, under the direction of a qualified medical physicist, which characterize the external therapy beams used for the patient treatments. This characterization should include, but is not limited to, the reference dose rate, output ratios, accessory attenuation factors, off-axis ratios, and depth dependence. A number of dosimetric systems for both photon and electron beams have been detailed;[22–25] however, physicists are encouraged to adopt the format and terminology recommended by AAPM Task Group 71 when their report becomes publicly available. An example calculation system that can be used for MUV is presented in Appendix A and is applied to clinically relevant examples in Appendix B. The equation notation presented in these appendices adheres as closely as possible to the draft recommendations of TG-71. A summary description of this notation is included in Table IV for reference. As with computer-based MUV systems, discussed below, a periodic review of dosimetric data in circulation for performing calculations should be undertaken to ensure the appropriateness and self-consistency of the data in use.

V.D. Computer-based MU verification programs

Most computer-based MU verification programs use an automated table look-up method similar to that outlined in Appendix A. Some more complex MU calculation programs use pencil beam or convolution/superposition algorithms based on the empirical data. For MU verification programs that use a series of piecewise equations for a depth dose curve, the MU calculation should be checked during commissioning at multiple points that fall within each piece of the parametrization. These computer programs require periodic QA to verify the continued data integrity and calculation algorithm functionality.

V.E. Verification TPS

Some institutions may use a second independent planning system for verification calculations. In this situation, identical beams and MU settings as in the primary plan should be

TABLE IV. TG-71 recommended symbol notation and definitions.

Equation symbol	Calculation type	Parameter description
CF	Photon, electron, arc	A dosimetric correction factor accounting for patient geometry effects. Most commonly used to characterize tissue inhomogeneity corrections, but can also include missing tissue and beam obliquity effects.
d	Photon, electron	Physical depth of the point of calculation, defined as the projection onto the central axis of the ray-line distance through the patient surface to the point of calculation.
d_{eff}	Photon, electron	Radiological depth to the point of calculation, scaled by the ratio of the physical density of the medium to water along the ray-line path length.
d_m	Photon, electron	The depth of maximum dose on the central axis.
d_0	Photon, electron	Normalization depth for the dosimetric system; the depth at which depth sensitive parameters take on a value of 1 or 100%. For electron fields, d_0 is set to d_m.
D_m	Photon, electron, arc	The depth of the maximum dose observed in a photon or electron depth dose curve.
D	Photon, electron, arc	The dose to be delivered to the point at a depth of d.
D_0'	Photon, electron, arc	The dose rate at the normalization depth of d_0 for the reference field size r_0. Typically set to 1 cGy/MU.
$F_{\text{air}}(r,\text{SSD})$	Electron	The air-gap correction factor that parametrizes the deviation from pure inverse square law behavior for an electron field.
g	Electron	The measured distance or gap between the patient surface in the center of the field and the standard treatment distance for an electron applicator.
IL_{Rx}	Arc	The percentage isodose line corresponding to the prescribed dose TD_{Rx} in the distribution.
MU	Photon, electron, arc	The counter setting for a treatment unit used to deliver the prescription dose. May refer to time for external beam treatment units using radioactive materials.
N	Arc	The number of arcs used in an arc plan.
$\text{OAR}(d,x)$	Photon	The off-axis ratio, defined as the ratio of the dose at a depth d for a point displaced by a distance x from the central beam axis measured at the isocenter plane, to the dose on the central axis for the same depth, machine output and scattering conditions.
$\text{PDD}(d,r,\text{SSD})$	Photon, electron	The percentage depth dose, defined as the ratio of the dose at depth d to the dose at d_m in water for a field size r and source to surface distance of SSD.
$\text{PDD}_N(d,r,\text{SSD})$	Photon, electron	The normalized percentage depth dose, defined as the ratio of the dose at depth d to the dose at d_0 in water for a field size r and source to surface distance of SSD.
r	Photon, electron	The field size measured at the surface of the patient.
r_c	Photon, electron, arc	The field size in the plane of the isocenter defined by the variable aperture collimation system closest to the radiation source. Typically defined by the secondary jaws or MLC system replacing the secondary jaws.
r_d	Photon, arc	The equivalent square of the treatment aperture defined by all collimating devices projected to the plane normal to the central axis containing the point of calculation at a depth d.
r_0	Photon, electron, arc	The reference field size for the treatment modality that defines the reference dose rate D_0'.

TABLE IV. (*Continued.*)

Equation symbol	Calculation type	Parameter description
$S_c(r_c)$	Photon, arc	The in-air output ratio for a field size r_c to the normalization field size r_0, typically defined by the positional collimating system closest to the radiation source.
$S_e(r_a,r)$	Electron	The relative in-phantom dose for a field size r, within an applicator of size r_a, at a depth of d_0 (typically close to d_m), normalized to the dose for the reference field r_0 within the reference applicator at the standard treatment SSD.
$S_p(r_d)$	Photon, arc	The ratio of the total dose in phantom at the center of the field for a field size of r_d to the total dose for the reference field size r_0 at a depth of d_0, for the same central axis energy fluence in both cases.
SPD	Photon	The source to point distance, defined as the projection of the ray-line from the source to the point of calculation onto the beam central axis.
SSD_0	Photon, electron, arc	The source to phantom surface distance for the normalization conditions and reference dose rate.
SSD	Photon, electron	The source to phantom or patient surface distance for the point of calculation, defined as the normal distance from the source to the plane orthogonal to the central axis that contains the surface intersection point of the ray-line from the source to the calculation point.
SSD_{eff}	Electron	The effective dosimetric position of the electron source for which the inverse square law behavior holds over the range of clinical useful displacements from SSD_0.
TF	Photon, arc	The attenuating tray factor for a photon beam, defined as the ratio of the radiation field output with the tray in place to that of the open field of the same field size and quality.
TD_{Rx}	Arc	The prescribed daily tumor dose.
$TPR(d,r_d)$	Photon	The tissue phantom ratio for an aperture defined equivalent square field size r_d at a depth of d.
$\overline{TPR}(\overline{d},r_d)$	Arc	The average TPR for an arced photon field size of r_d, calculated from an average of TPRs for the depths sampled at intervals along the full arc path.
$WF(d,r,x)$	Photon	The attenuation wedge factor for a depth d and exposed field size in the isocenter plane of r at an off-axis point x. Typically, the small reduction in field size resulting for tertiary shielding (MLC, cerrobend shielding) can be neglected so that $r=r_c$. In older dosimetry systems, this factor has been presented as a composite of central axis and off-axis factors; i.e., $WF(d,r,x)\equiv WF(d,r,0)\cdot WOAR(d,x)$
$WOAR(d,x)$	Photon	The wedged off-axis ratio, defined as the ratio of the dose at a depth d for a point displaced from the central axis by a radial distance x measured in the isocenter plane to the dose at the same point off-axis in an open field of the same dimensions which delivers the same central axis dose as the wedged field. (Obsolete factor. See WF.)
x	Photon	The off-axis distance, defined as the distance from the central axis to a fan line through the point of calculation, measured in the isocenter plane.

used in the verification plan and the patient geometry used in the verification should be derived from the same patient image source as the primary calculation. A comparison point common to both systems should be selected according to the criteria discussed in Sec. V A and used to determine the level of dosimetric agreement between the two systems.

V.F. Verification by measurement

Measurement confirmation of the dose delivered to a point can be used as a verification method; however, measurement has several drawbacks as a routine verification tool for nonmodulated fields. First, it can be difficult to ad-

equately replicate the patient geometry, particularly for nonflat surfaces or fields with flash. Geometrical replication becomes even more difficult when heterogeneities need to be taken into account. Second, measurements are time-consuming. For the simple geometrical conditions that measurements usually use, it is easier and faster to perform a calculation, and under these conditions, the accuracy of a basic manual calculation should be comparable to the measurement accuracy.

Measurement is most useful when the difference between the primary and the verification calculations is outside the action levels. In this case, measurement can be used to help determine the cause of the disagreement. Section VI B further discusses this use of measurement. In addition, *in vivo* verification measurement on the first day of treatment is recommended for specialized treatment procedures, such as total body irradiation and total skin electron treatment, which are typically not planned with TPS but rather use a simple hand or computerized calculation to determine MU.

V.G. Stereotactic therapy

Stereotactic radiosurgery and radiotherapy (SRS and SRT) are highly conformal small target volume techniques that are realized through a diversity of technical solutions ranging from arced fields on a conventional linear accelerator to specialized treatment units such as CyberKnife and GammaKnife. The technique selected for performing a MUV will depend heavily on the technical details of the radiation delivery process. The corresponding action levels need to take into account any limitations of the MUV method, including difficulties associated with small field dosimetry and the dosimetric requirements of the particular treatment. For GammaKnife and CyberKnife, the AAPM has two active task groups that specifically address these specialized treatment units (TG 178 on Gamma Stereotactic Radiosurgery Dosimetry and Quality Assurance and TG 135 on QA for Robotic Radiosurgery), and the reader is referred to these reports when they become publicly available.

For linac-based SRS/SRT treatments using uniform arcs or multiple uniform fields on a target large enough to establish electron equilibrium at the calculation point, the verification techniques and action levels in this report can be applied. For smaller fields, the user should have available measured data, e.g., from commissioning, that can be used in the verification. The reports of TG 155 on Small Fields and Non-equilibrium Condition Photon Beam Dosimetry, when they become publicly available, and TG 101 on Stereotactic Body Radiotherapy[26] will provide useful references.

VI. ACTION LEVELS AND REMEDIAL ACTIONS

VI.A. Action levels for MU disagreement

Total uncertainty in absolute delivered dose of no more than 5% is a widely accepted goal for effective radiation treatments.[1,3,27] One purpose of the independent MU verification process is to help assure that this goal is achieved for at least one representative point within the target volume. While AAPM Task Group 53 (Ref. 5) and others[28] gave recommended criteria for the absolute accuracy of the dose calculation, the literature on the expected level of agreement between primary and verification calculations for modern image-based 3D planning systems is limited.[29–34] The action level guidelines given in Tables II and III are therefore based primarily on the collective experience and expectations of the task group members. A base action level of 2% was postulated for simple field geometries, consistent with the TG-53 criterion of 2% dose accuracy between calculations and measurements. From this starting point, additional range was added to the level to account for the increased uncertainties of complex treatment geometries that deviate from a uniform irradiation of the central region of a rectangular field incident on a flat-surfaced homogeneous volume.

These guidelines are consistent with and inclusive of the published studies of verification calculation agreement. Task Group 40 (Ref. 1) recommended agreement within 2% in many circumstances, but indicated that 5% is more realistic if sophisticated algorithms are used with substantial field blocking or significant heterogeneity correction, or if electron beams are used. Leszczynski and Dunscombe[29] used a spreadsheet program that employed radiological path length density corrections to verify calculations from a commercial 3D TPS. For all eight sites analyzed, the average ratio of verification MU to TPS MU was 1.002, with standard deviation of 0.012. The largest variations were seen for supraclavicular fields (average=1.013, σ=0.010) and rectal fields (average=0.990, σ=0.009). However, they did not include an off-axis factor for the supraclavicular fields nor did the spreadsheet calculation account for beam hardening due to the physical wedge for the rectal fields. Chan *et al.*[30] compared manual calculations to homogeneous calculations from a 3D TPS with the aim of evaluating the usefulness of the manual calculation as a verification tool. They found average ratios of verification to TPS MUs for four sites ranged between 1.010 and 1.013, and standard deviations of 0.005–0.016, with the largest variations encountered for breast treatments and supraclavicular fields. They analyzed the supraclavicular fields by systematically reducing the calculation complexity (removing wedges, flattening out the surface, etc.) and found increasing agreement with decreasing complexity. They concluded that the largest single factor influencing the agreement between their verification and TPS calculations was surface/contour irregularity, but there are other contributing factors as well. Prado *et al.*,[31] Ayyangar,[32] Kay and Dunscombe,[33] and Kay and Meyer[34] all proposed correction methods to better account for surface shape and scattering characteristics in hand calculations for breast tangents. With their corrections, they quoted 0.1%–2.3% average differences between verification and 3D TPS calculations.

Guidelines for action levels for disagreement between primary and verification MU calculations are given in Tables II and III in terms of percentage differences. For low dose, low MU fields, it may be more reasonable to use an absolute difference criterion of 1–3 MU (or 1–3 cGy for dose com-

parisons), since for small total MU, a difference of a few MU can result in a large percentage difference but negligible clinical consequences. ***The action levels provided are only guidelines and are not to be used as goals or requirements for any particular situation***. The particulars of the dose calculation and verification methodology at each institution, as well as particular patient circumstances, may affect these expectations. ***The radiation oncology physicist at each institution should evaluate the expectations for each circumstance to determine relevant criteria for verification MU agreement for that institution***. Different criteria may be used for different treatment sites or techniques. The physicist should also consider the clinical significance of each field to the overall treatment.

The action level guidelines are divided into two tables, depending on whether or not tissue heterogeneities are taken into account in the primary calculation. Table II applies when the patient density is assumed to be homogeneous, while Table III applies when heterogeneity calculations are performed. The task group strongly recommends that if the primary calculation applies heterogeneity corrections, then the verification calculation should also, although the methods for heterogeneity correction may, and typically do, differ. Further discussion of heterogeneity-corrected calculations can be found in Appendix A 1 a.

In order to obtain a valid verification calculation, the limitations of both primary and verification calculation algorithms must be recognized in the selection of the point of comparison. The action level tables assume that either the guidelines discussed in Sec. V A have been followed or corrections are applied to account for disequilibrium effects.

These action level guidelines apply for conventional electron beam as well as all non-intensity modulated photon beam calculations, including arcs and stereotactic fields. In each case, the physicist should evaluate the complexity of the field and calculation and determine the appropriate value to use based on the table entries.

Appropriate care should be exercised in applying these guidelines to the field-in-field (FIF) treatment beams. The FIF technique incorporates into a single field a limited number of subfields that are usually forward-planned rather than derived from computerized optimization. Some institutions consider FIF a non-IMRT technique, while others classify it as IMRT. The MU verification for these types of fields should be performed accordingly. For those institutions verifying these fields through calculations, it needs to be recognized that it may be difficult to define a single verification point that is adequate for all the subfields. Each center will need to determine an acceptable minimum subfield sampling for the calculation point and under what conditions a second calculation point may be considered.

Explanation of table format. A number of factors will affect the achievable agreement range between primary and verification calculations for typical treatments at each institution. In general, algorithms applying the same methodology would be expected to achieve a closer level of agreement, while disparate algorithms (e.g., pencil beam vs convolution, or convolution vs Monte Carlo) would be expected to require a wider acceptance range. This difference is reflected in the tables by the division of the columns into these two broad categories. The manner in which patient geometry information is used by the respective calculations influences action levels as well. The more similarity that exists in the handling of patient geometry, the closer the level of agreement expected. If both calculations use the same patient model, whether it is a full CT data set or a uniform cube phantom approximation, then a better agreement is expected than when one calculation uses the full CT data set and the other uses a uniform cube phantom approximation. An intermediate patient model between these two extremes is also listed in the tables, referred to as "approximated patient geometry," and is intended for calculations that apply corrections to a uniform cube phantom geometry. Corrections for such conditions as tangentially incident fields, heterogeneity calculations, or irregular surfaces would apply to this category of patient geometry modeling. For heterogeneity-corrected calculations, the uniform cube phantom approximation is no longer sufficient to achieve an acceptable confidence level for MU verification, so that column has been removed from Table III.

Field complexity also influences expected agreement between primary and verification calculations and is reflected in the rows of Tables II and III, which are arranged roughly in terms of increasing field complexity. "Contour change" indicates situations with an oblique entrance or tangential beam geometry, as well as surface irregularity. The examples in Appendix B, which are summarized in Table V, illustrate the application of these guidelines to a variety of clinical cases.

VI.B. Remediation of MU discrepancies

Verification calculations that results in a variance from the primary MU calculation outside the expected range for the conditions of the calculation (see Sec. VI A) need to be investigated to ensure that the primary calculation has been carried out to an acceptable accuracy. Thus, when a discrepancy is noted, the first action of the physicist reviewing the calculation should be to verify that a calculation error has not been made (see Sec. II B and Table I for a list of common errors).

The investigation into a discrepancy should start with the review and comparison of parameters used in the two calculations. This review should be sufficiently thorough to confirm that the correct parameters, such as energy, field size and depth, were used, and that all accessories have been properly taken into account. Particular attention should be devoted to ensuring that both calculations correctly interpret the treatment prescription in a consistent manner. The magnitude of the discrepancy should serve as an indicator of the cause; a very large difference would likely result from an improper accounting of an attenuator such as a wedge, while moderate differences could result from improperly accounting for the effects of block trays or bolus applied to the patient. A large difference could also indicate a misinterpre-

TABLE V. Clinical examples presented in Appendix B.

Case	Calculation geometry	Same/different algorithm	Patient geometry	Within action levels?
Brain	Homogeneous; minimal shaping	Different	Approx, patient geometry; flash modeled as block	Yes
Breast	Homogeneous; off-axis wedge	Different	Uniform cube phantom	No
			Approx. patient geometry; scatter correction factor	Yes
			Approx. patient geometry; flash modeled as block	Yes
Mantle	Heterogeneous; large field	Different	Approx. patient geometry; radiological depth	Yes
Modified mantle	Heterogeneous; large field	Different	Approx. patient geometry; Day's method	No
			Approx. patient geometry; additional scatter correction factor	Yes
Large field lung	Heterogeneous; large field	Different	Uniform cube phantom (homogeneous)	No
			Approx. patient geometry; radiological depth	Yes
Small field lung	Heterogeneous; small field, low-density heterogeneity	Different	Approx. patient geometry; radiological depth	No
			Approx. patient geometry; O'Connor's theorem	No
			Approx. patient geometry; effect of calculation point position	Yes
Electron distribution	Homogeneous, minimal field shaping	Different	Uniform cube phantom	Yes

tation of the prescribed dose or a misuse of the inverse square law (ISL). This minimal investigation typically occurs during plan review as part of the routine planning QA, but such inputs to the MUV should be carefully scrutinized for accuracy and physical soundness in the context of the patient geometry when the MUV is first generated.

If this basic review fails to identify the cause of an actionable discrepancy, the next step should be to confirm that an appropriate comparison point has been chosen. The guidelines for comparison point placement presented in Sec. V A should be followed where possible. In situations where point placement has been identified as an issue, new calculations for a more appropriately positioned point should be performed.

Differences in accounting for patient geometry between the primary and the verification calculations can also lead to large discrepancies between results. For example, the uniform cube phantom approximation used in many verification calculations tends to overestimate the contribution of scatter to the point of calculation in situations such as tangential opposed beams for breast treatment. Methods such as described in Appendix A 1 b and illustrated in the examples in Appendix B should be applied to the verification calculation to better estimate the scatter contribution.

Density corrections are required for verification of calculations which include heterogeneity effects (see Appendix A 1 a). The verification calculation must at least take into account the radiological thickness of tissues overlying the point of calculation. Point placement is crucial in obtaining agreement within an acceptable range, and care must be taken that the ray-line path from the source to the calculation point provides a reasonable estimate of conditions experienced by the calculation point. An example of where this would be violated is a situation where the ray-line passes tangentially through the cortex of a long bone and, consequently, overestimates the impact of the heterogeneity. Such conditions are typically identified during planning review, but may not be identified until the review of the verification calculation.

Heterogeneity calculations involving lung tissue tend to be among the most difficult for achieving an acceptable verification calculation. A thorough understanding by the responsible physicist of the capabilities and limitations of both the primary and the verification MU calculations is required for both setting the action levels at a clinic and for resolving discrepancies between calculations under these conditions. This knowledge will aid the physicist in determining to what level radiation transport processes are modeled in the MUV in comparison to the primary calculation, and how additional corrections such as discussed in Appendix A 1 b and applied in the examples of Appendices B 4–B 6 may be used to confirm algorithmic differences as the source of observed discrepancies. At a minimum, if a discrepancy is attributed to differences in the calculation algorithms, an assessment to confirm that the discrepancy is the correct order of magnitude and direction should be made. Small field calculations within lung can be particularly difficult to achieve within the desired accuracy since they experience effects from electronic disequilibrium.[35] In order to confirm that a discrep-

ancy between primary and verification calculations arises from differences in the heterogeneity corrections and not some other causes, a comparison between calculations both using a homogeneous patient geometry may be performed to confirm this cause. While such a calculation does not serve to validate the treatment MU setting, it does serve to identify the source of the discrepancy. A complete verification of the MU setting in this situation would still require the derivation and application of a correction factor (CF) applicable to the patient geometry and perhaps a refinement of point positioning (see the example in Appendix B 6 c).

The possibility of incorrect data or modeling in either the primary or the verification dosimetry system cannot be dismissed. No matter how thorough the commissioning or QA, it is not possible to completely verify a calculation system under every possible situation. In addition, data in TPSs can become corrupted, either through system failure or human error. Parameters used for the calculation should be checked against data tables for consistency with standard data and reasonable data trends.

It is recognized that in some instances, even additional correction methods may not be able to produce a verification calculation within the expected level of agreement. In this situation, it may be necessary to resort to a phantom measurement to provide a verification of the treatment MU setting. In geometries with complex surface shape or heterogeneities, an accurate measurement may not be practical or even possible. In this situation, TPS commissioning tests performed by the institution or reported in literature may be useful to quantify the expected effects and provide guidance on the direction and magnitude of the observed discrepancy. When performing measurements to validate a calculation, it is also important to consider and investigate treatment unit issues as a possible source of discrepancy.

In situations where the same geometry is encountered in multiple patients, technique-based corrections factors, derived either from the literature (see example in Appendix B 2) or from measurements, may be employed; however, disagreements outside the action levels can result from conditions, which deviate from the reference conditions, and such a situation may need to be investigated through physical measurements. Measurement, however, is often not the entire solution since discrepancies may result from complexity in the patient geometry, which may not be reproduced by the measurement geometry.

Inevitably, there will be circumstances under which all methods of calculation and measurement fail to achieve an acceptable level of agreement between the primary treatment and verification calculations. Such behavior may indicate that the chosen method of treatment results in a large dosimetric uncertainty for the treatment in question and, consequently, another technique or treatment approach might be considered. In situations where it is judged that the planned treatment must be delivered, then the physicist should inform the oncologist that the planned treatment has a larger than normal dosimetric uncertainty and should also report the most likely cause of the uncertainty. This uncertainty does not necessarily infer an error in treatment delivery; the discrepancy between primary and verification calculation results in many cases may not be due to inaccuracies in the primary calculation but to limitations of the verification system. When large unresolved discrepancies are routinely encountered for a particular type of treatment, then a detailed investigation should be performed to clearly identify the cause of the discrepancy and derive correction factors to accurately account for the effect.

The resolution of discrepancies between the primary MU and verification MU calculations requires an accurate description of the radiation environment in the patient at the point of calculation. This information typically is provided by the standard plan documentation generated by the TPS. It is imperative that this information be provided in a format that permits a physicist to identify the type and magnitude of the individual corrections applied in the dose calculation and to assess the appropriateness of these factors for the known geometry of the patient. Ideally, this information should be provided in the form of standard parameters that can be used in a simple hand calculation, such as SSD, depth, radiological depth, off-axis distance, and equivalent square. The task group strongly recommends that commercial TPS vendors make this information readily available in their plan summary documentation. Where such treatment parameters are provided, it is also important that user documentation explicitly states the distance at which these parameters are defined (i.e., SSD, SAD, or point of calculation). An assessment of the TPS plan summary documentation for these parameters is an important consideration in the selection of a TPS for clinical use.

VII. SUMMARY

The MUV is the independent verification overseen by a qualified medical physicist that validates that the appropriate machine settings are used to deliver the intended dose within the patient. The MUV remains an important element of the radiation therapy QA program that ensures a safe and accurate patient treatment.[4,11,12,36] It is usually performed as part of the pretreatment plan check, and while it is an important component of that check, it should be recognized that it is only one component and does not alone constitute nor take the place of a comprehensive review of the plan. In addition, continuing education programs for all staff involved in treatment planning and delivery, including physicians, therapists, dosimetrists, and physicists, should include training to explain the MUV calculation and the adopted institution action levels under various treatment conditions.

A key aspect of the MUV that ensures its continued value as a QA tool is the independent nature of the check. Despite the increase in treatment complexity that causes a reliance on the primary TPS for the determination of some of the treatment parameters, the verification can and should be independent in several key aspects. First, a calculation program and/or methodology that is separate from that of the primary calculation should be used. Second, the beam data and treatment parameter files should be separate and independent of those used by the primary TPS. Finally, when the electronic

transfer of patient specific parameters from the primary to the verification system is used, the physicist should assess and evaluate the suitability of those parameters by an independent means, which, for example, may include simple estimation of an aperture equivalent square from a BEV, or comparison of physical and radiological depths to ranges typical of the anatomical site based on clinical experience. The establishment and maintenance of an independent MUV program is an important responsibility of the medical physicist.

A variety of methods can be used to perform the MUV, as described in Sec. V and Appendix A. All software programs used for the verification calculation should be thoroughly commissioned and tested prior to use, as should the primary TPS. It is important that the physicist knows the accuracy and limitations of both the primary and the verification systems in order to set reasonable and achievable action levels and to better interpret the causes of differences between the two results.

The level of agreement achievable between primary and verification calculations depends on the details of the patient geometry, the primary and the verification calculation programs, and the clinical situation, in addition to whether corrections for tissue heterogeneities are used. It is therefore reasonable and expected to have different action levels for different situations. Each institution must determine the proper action levels for that particular clinic. Results from planning system commissioning are useful in establishing these levels. Tables II and III may be used as guidelines, but should not be adopted without evaluation. The action levels established by the institution should be documented in written protocols and disseminated to staff through appropriate continuing education activities. With appropriate action levels and responses, the MU verification is both an effective and efficient QA tool that can help identify and reduce treatment errors.

APPENDIX A: MANUAL CALCULATION METHODOLOGIES

1. Manual photon calculations

The basis of any MU calculation system used for the verification of the delivered dose is the derivation of a dose (or dose rate) delivered in patient from the known dose rate under the reference conditions. A typical methodology for performing this calculation for a simple water equivalent slab phantom geometry is illustrated in Fig. 1 using a conventional manual calculation system.[22,24,37] Dose to point P is a product of the reference dose rate, number of MU used, in-air output ratio (S_c), phantom scatter factor (S_p), the TPR, and the distance correction factor. As shown schematically in Fig. 1, the dose to a point P in a flat-surfaced water phantom can be calculated using the equation,

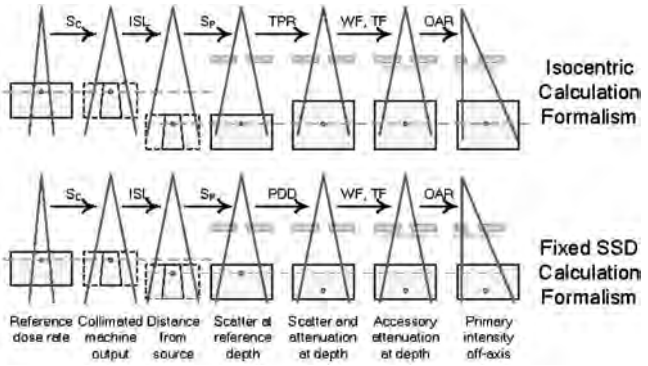


FIG. 1. Development of Eqs. (A2) and (A3) for photon dose calculation to a point in a water equivalent slab phantom. Successive application of factors transforms the dose rate from the reference condition to the conditions of treatment geometry and depth. Steps in the calculation where the phantom scatter differs from the irradiating field size during the application of S_c and ISL factors are indicated by sectioning the phantom, where the portion enclosed by the solid lines signifies the inclusion of reference field size phantom scatter. The solid circle represents the position of the calculation point, while the dashed line crossing several phantoms indicates changes in the point distance from the source.

$$D = \mathrm{MU} \cdot D_0' \cdot S_c(r_c) \cdot S_p(r_d) \cdot \mathrm{TPR}(d, r_d) \cdot \left(\frac{\mathrm{SSD}_0 + d_0}{\mathrm{SPD}}\right)^2 \cdot \mathrm{OAR}(d,x) \cdot \mathrm{TF} \cdot \mathrm{WF}(d,r,x). \quad \text{(A1)}$$

The parameters used in this and other equations follow the draft recommendations of AAPM Task Group 71 and are defined in Table IV. TPSs do not always use these definitions, particularly for off-axis ratio (OAR) distance and depth; users must understand how parameters are defined within their TPS.

As shown in Fig. 1, the product of MUs and the normalization dose rate (dose rate at the normalization point for the reference field) will yield the dose at the normalization point for the reference field size. However, calculation conditions are typically different from the reference conditions. The next three terms correct for the changes in output, attenuation, and scatter conditions from the reference conditions. The in-air output ratio, S_c, also called the collimator scatter factor, corrects for treatment unit output changes with field size. It is typically a complex function of the field size projected to the isocenter and depends on the design of the treatment unit collimation system.[38–41] The phantom scatter factor, S_p, and the TPR together correct for differences from the reference conditions in scatter and attenuation, with S_p primarily accounting for changes in scatter and TPR accounting for the effects of tissue attenuation as well as changes in scattering conditions with depth. These factors depend on the effective equivalent square[42–44] of the irradiated area (at SPD), which is typically estimated from the field shape defined by the MLC or poured blocks. The ISL term accounts for the intensity change resulting from source-to-point distance differences between the calculation and reference positions. The OAR also accounts for changes in intensity, but in this case resulting from changes in beam characteristics

away from the central axis. As these changes include both beam quality and intensity, this factor is a function of off-axis distance and depth.

The factors TF and WF account for the attenuation from beam modulators, shaping devices, and patient support devices in the beam path required for typical treatment plans. Thin plastic trays placed into the treatment beam to support field-shaping blocks typically reduce the beam intensity by a few percent and are frequently characterized by a single factor, TF, depending only on beam energy for all points in the field. The behavior of the wedge factor, WF, on treatment settings is closely tied to the design of the treatment unit, particularly the position of the wedge relative to the secondary and tertiary collimation system.[45–48] This factor primarily accounts for the changes in intensity arising along a ray-line path to the point of calculation due to either the thickness of material or the change in the jaw position during irradiation in the case of a virtual wedge, and as such exhibits large variations with off-axis position x. The WF typically accounts for field size, depth, and SSD dependent effects[49–51] arising from scatter radiation emanating from the wedge filter itself.

Monitor unit calculations for isocentric beams can be performed by rearranging Eq. (A1) into the form,

$$\mathrm{MU} = \frac{D}{D_0' \cdot S_c(r_c) \cdot S_p(r_d) \cdot \mathrm{TPR}(d,r_d) \cdot \mathrm{OAR}(d,x) \cdot \mathrm{TF} \cdot \mathrm{WF}(d,r,x) \cdot \mathrm{CF}} \cdot \left(\frac{\mathrm{SPD}}{\mathrm{SSD}_0 + d_0}\right)^2. \tag{A2}$$

Additional corrections for specific patient geometry may be incorporated through the CF factor to account for missing tissue in tangential geometries, sloping surfaces, and tissue heterogeneities.

Monitor unit calculations for nonisocentric en face beams can also be done using the percent depth dose values according to Eq. (A3),

$$\mathrm{MU} = \frac{D \cdot 100\%}{D_0' \cdot S_c(r_c) \cdot S_p(r_{d_0}) \cdot \mathrm{PDD}_N(d,r_{d_0},\mathrm{SSD}) \cdot \mathrm{OAR}(d,x) \cdot \mathrm{TF} \cdot \mathrm{WF}(d,r,x) \cdot CF} \cdot \left(\frac{\mathrm{SSD} + d_0}{\mathrm{SSD}_0 + d_0}\right)^2, \tag{A3}$$

where the TPR term in Eq. (A2) has been replaced by a percentage depth dose term appropriate to the exposed field. This also affects the ISL term compared to Eq. (A2), which now accounts only for that part of the change in intensity due to a difference between the calibration and treatment SSDs.

a. Heterogeneity corrections

The heterogeneity correction is usually small in most clinical sites, such as breast or prostate, but can be substantial for chest treatments when a large volume of lung is being irradiated or when the tumor is surrounded by lung tissue. Typically, heterogeneity corrections will improve the dose accuracy compared to a homogeneous dose calculation.

For a heterogeneous dose calculation, the most important parameter is the radiological depth along the ray-line to the point of calculation. While the radiological depth is typically the largest component for this correction, in low-density regions, such as the lung, electronic disequilibrium effects due to the lateral extent of the field and rebuild-up can also be significant.[20,52] Care must be taken to avoid selecting a calculation point near the tissue-lung interface to avoid these effects due to secondary electron transport.

The inhomogeneity CF is defined as

$$\mathrm{CF} = \left(\frac{\text{dose in heterogeneous medium}}{\text{dose at same point in homogeneous medium}}\right). \tag{A4}$$

The ratio of TAR (RTAR) method is based on the radiological-path-length scaling theorem proposed by O'Connor.[53,54] According to this theorem, particles in a medium of half the density will travel twice as far to undergo the same number of interactions. Using more conventional parameters, the RTAR inhomogeneity correction factor may be calculated using the equation

$$\mathrm{CF} = \left(\frac{\mathrm{TPR}(d_{\mathrm{eff}},r_d)}{\mathrm{TPR}(d,r_d)}\right), \tag{A5}$$

where the peak scatter factor term divides out in the ratio, leaving only a ratio of TPR terms. The RTAR method calculates the dose contribution from the primary beam accurately but the scatter contribution is inaccurate because the size, shape, and location of the heterogeneous medium are not included in the scatter integration. RTAR assumes that the heterogeneous medium is infinite in the lateral dimensions (infinite slab approximation). Batho[55] proposed a method that was later extended by Sontag and Cunningham,[56] based on the radiological path length scaling for the primary dose, expressed as

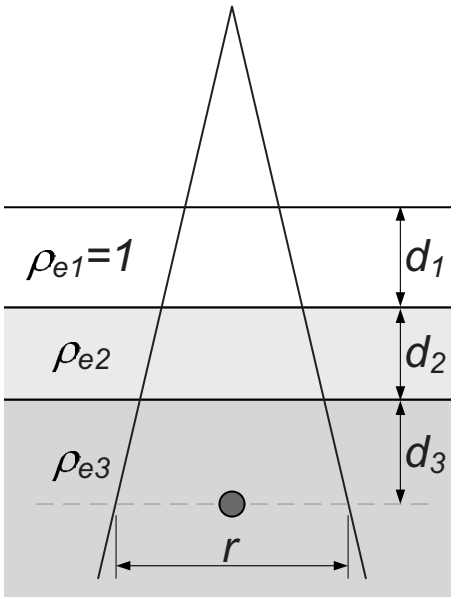


FIG. 2. Photon dose calculation to a point in a heterogeneous phantom. The first layer of material is assumed to be water equivalent.

$$\mathrm{CF} = \left(\frac{\mathrm{TPR}(d_3, r_d)^{\rho_{e3}-\rho_{e2}}}{\mathrm{TPR}(d_2 + d_3, r_d)^{1-\rho_{e2}}} \right), \tag{A6}$$

where distances d and electron densities ρ are defined in Fig. 2. These simple ratio methods described above do not take into account the effect of the lateral dimension of the heterogeneity. To characterize the impact of the size of an inhomogeneity on scatter, an effective equivalent square can be used.[57] It is usually determined as the electron density scaled lateral dimension, ρ_r. While the effective square can be calculated through an equivalent TAR method for TPS calculations, it is rare for it to be used in MUV calculation. For more information on these methods, refer to AAPM Report No. 85 from Task Group 65.[20]

In addition to calculation methods, the quality of relative electron density information in the planning CT needs to be considered when employing heterogeneity corrections. Nontissue high-Z materials can be introduced into the patient either through prosthetic materials implanted in the patient or the use of CT contrast to help visualize target regions. For patients with high-Z material implants such as hip prostheses within the treatment field, the reader is referred to the AAPM Task Group 63 report for guidance in properly accounting for the dosimetric effects.[58] The same techniques may also be used for other materials that exhibit an anomalously high density, such as dental amalgam and bone cement. The use of radio-opaque contrast during CT simulation can also introduce anomalously high-density regions into the planning CT, which will not be present during treatment. While several clinical investigations have found that such regions in the vicinity of the target introduce only a small dose perturbation into the calculation,[59–64] instances where it is judged to have a significant effect can be addressed by contouring the region and setting the relative electron density to that of the sur-

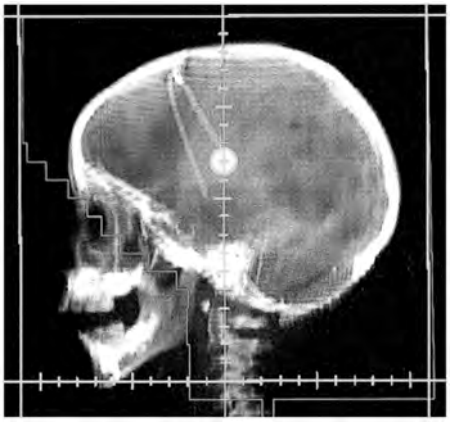
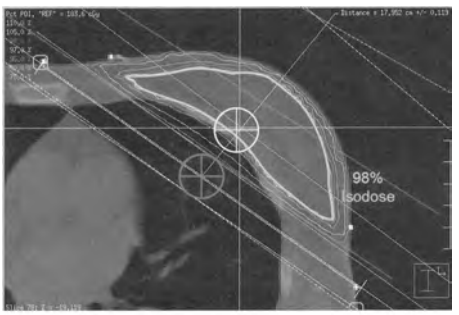
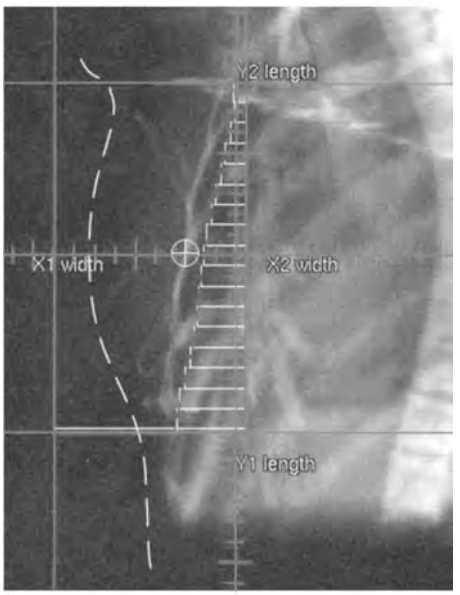


FIG. 3. BEV showing field blocking used to estimate the exposed equivalent field size for the brain field MUV in Appendix B 1. The calculation point position in the field is indicated by the circle.



(a)



(b)

FIG. 4. Geometry for standard breast MUV calculation in Appendix B 2. (a) Placement of the calculation point within the patient in soft tissue and away from the tissue interfaces. (b) BEV for the field used to estimate the missing scatter from the tangential field. The dashed white line indicates the position of the breast horizon and the crossed circle indicates the location of the calculation point in the field.

TABLE VI. Whole brain case.

Reference conditions		Treatment description		Dosimetric parameters	
SAD	100 cm	D	75 cGy	r_c equivalent square	21.0 cm
SSD_0	98.5 cm	SPD	100 cm	r_d equivalent square	15 cm
d_0	1.5 cm	d	7.6 cm	$S_c(r_c)$	1.033
r_0	10×10 cm^2	r_c	22×20 cm^2	$S_p(r_d)$	1.009
D_0'	1 cGy/MU	x	12 cm	TPR(d,r_d)	0.878
		Blocked field area[a]	29%	OAR(d,x)	1.042

[a]Estimated from BEV shown in Fig. 3.

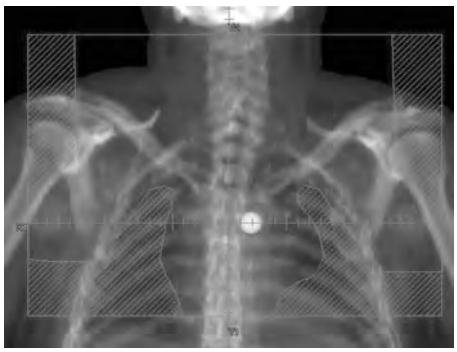


FIG. 5. BEV of the anterior upper mantle field calculated in Appendix B 3. The circle shows the prescription point used in the MUV calculation, which has been located in the middle of the mediastinum, well away from low-density tissue interfaces.

rounding tissue.[20] In these instances, the best estimate of regional relative electron density should be used for both the primary and the MUV calculations.

b. Scatter corrections

Corrections in MUV calculations are typically required to account for changes in the scattered radiation from uniform cube phantom geometry arising from patient and beam geometry effects. Correction methods of this type can typically be expressed through a change in the field size,

$$\mathrm{CF}=\left(\frac{S_p(r_d')\cdot\mathrm{TPR}(d,r_d')}{S_p(r_d)\cdot\mathrm{TPR}(d,r_d)}\right), \tag{A7}$$

where r_d' is the effective field size. A number of different methodologies for calculating the effective field size have been reported for specific geometries such as breast tangents.[31,33,34] Simpler methodologies using this approach have also been used for tangential fields treating areas such as larynx and limbs. For these corrections, an equivalent field size is obtained by considering the area of missing tissue seen in the BEV as though the region were under a shaped block or by estimating an average exposed field width from the observed BEV horizon. More complex scatter corrections may be required for special circumstances such as arise when field attenuators are used or in reduced scatter conditions such as mediastinal treatments irradiating a significant portion of lung. In these cases, a more careful assessment of the scatter may be obtained using Day's method[65,66] to estimate the reduced scatter from lower density material. In most clinical cases, the scatter correction in a photon field can account for a 2%–8% change in the MU value, but can be larger for small stereotactic fields.

c. Arc therapy

The formalism for an arc beam calculation differs subtly from a static beam. The formula

$$\mathrm{MU}=\frac{D}{\dot{D}_0\cdot S_C(r_C)\cdot S_p(r_C)\cdot\overline{\mathrm{TPR}}(d,r_C)} \tag{A8a}$$

is very similar to the more generalized equation (A2), with the elimination of accessory factors, and the modification of TPR. This formalism only supports a calculation to the isocenter of the arc. In this calculation, the depth to the isocenter is regularly sampled over the path of the arc and the mean TPR is calculated according to the equation,

$$\overline{\mathrm{TPR}}(d,r_C)=\frac{1}{n}\sum_i^n \mathrm{TPR}(d_i,r_C), \tag{A8b}$$

where the index i sums over the equally spaced segments along the arc (e.g., every 10°).[24]

The preceding formalism applies only to static arc delivery. A number of dynamic arc delivery techniques such as conformal arc therapy, intensity modulated arc therapy, volu-

TABLE VII. Wedged breast case.

Reference conditions		Treatment description		Dosimetric parameters					
				Uniform cube		Averaged geometry		Reduced scatter	
SAD	100 cm	D	91.9 cGy	r_d equivalent square	10.9 cm	r_d equivalent square	10.9 cm	r_d equivalent square	8.9 cm
SSD_0	98.5 cm	d	6.0 cm	$S_c(r_c)$	1.013	$S_c(r_c)$	1.013	$S_c(r_c)$	1.013
d_0	1.5 cm	r_c	10.5×17 cm^2	$S_p(r_d)$	1.004	$S_p(r_d)$	1.004	$S_p(r_d)$	0.990
r_0	10×10 cm^2	r_d	8×17 cm^2	TPR(d,r_d)	0.901	TPR(d,r_d)	0.901	TPR(d,r_d)	0.891
D_0'	1 cGy/MU	x	2.6 cm	WF(d,r,x)	0.660	WF(d,r,x)	0.660	WF(d,r,x)	0.660
		Missing scatter	−2 cm in field width	OAR(d,x)	1.020	OAR(d,x)	1.020	OAR(d,x)	1.020
						CF	0.96		

TABLE VIII. Mantle case.

Reference conditions		Treatment description		Dosimetric parameters	
SAD	100 cm	D	84.6 cGy	r_d equivalent square	19.4 cm
SSD_0	98.5 cm	d_{eff}	8.7 cm	$S_c(r_c)$	1.037
d_0	1.5 cm	r_c	36.4×25 cm^2	$S_p(r_d)$	1.028
r_0	10×10 cm^2	x	1.8 cm	TPR(d,r_d)	0.845
D_0'	1 cGy/MU	SPD	98.4 cm	OAR(d,x)	1.008
				TF	0.960

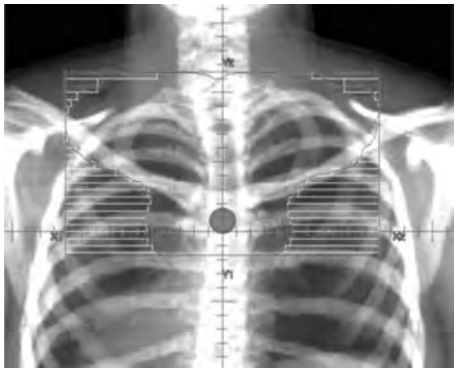


FIG. 6. BEV for the modified mantle field of Appendix B 4. The calculation point centered in the mediastinum is marked by the circle.

metric modulated arc therapy, and helical IMRT (tomotherapy) have been developed and have been implemented in a number of clinics. A consistent methodology for an independent MUV calculation of these modalities has yet to be developed.

2. Manual electron calculations

Electron beams are typically treated in en face geometry with custom field-defining apertures. The output of the electron beam depends on the beam energy, the SSD, the applicator/cone size, and the size of the field-defining aperture. Typically, electron beam output factors are measured at the depth of maximum dose for different field sizes and different SSDs. These measured values are used to compute the monitor units.

For regularly shaped large field sizes at standard SSD, one commonly used formula for the electron calculation is

$$\mathrm{MU} = \frac{D}{D_0' \cdot S_e(r_a,r)}. \tag{A9}$$

The output factor, S_e, for a rectangular field of $W\times L$ can be estimated using the following relationship:[67]

$$S_e(r_a,L\times W) = \sqrt{S_e(r_a,L\times L)\cdot S_e(r_a,W\times W)}. \tag{A10}$$

For nonstandard (extended) SSDs, there are two techniques to correct for the change in distance, g, from the reference conditions: The effective SSD method,

$$\mathrm{MU} = \frac{D}{D_0' \cdot S_e(r_a,r)} \cdot \left(\frac{\mathrm{SSD_{eff}} + d_0 + g}{\mathrm{SSD_{eff}} + d_0}\right)^2, \tag{A11}$$

and the air-gap parametrization method,

$$\mathrm{MU} = \frac{D}{D_0' \cdot S_e(r_a,r)\cdot f_{\mathrm{air}}(r,\mathrm{SSD})} \cdot \left(\frac{\mathrm{SSD_0} + d_0 + g}{\mathrm{SSD_0} + d_0}\right)^2. \tag{A12}$$

While these methods work well for typical regularly shaped fields, careful application is advised for irregular and small fields to ensure that loss of electron scatter does not introduce significant deviations in dose. In some cases, the output factor may be measured directly at the treatment distance and used directly in Eq. (A9); if a mixture of correction methods is employed, extreme care in the MU calculations should be exercised to ensure the effect of the extended distance is not applied twice. For more details on the formalism for dosimetric parameters and measurement techniques, refer to AAPM Task Groups 25,[68] 70,[69] and 71 report on MU calculations.

APPENDIX B: MONITOR UNIT VERIFICATION EXAMPLES

The following examples of verification calculations (summarized in Table V) were selected to illustrate the application

TABLE IX. Modified mantle case.

Reference conditions		Treatment description		Dosimetric parameters			
				Field shaping		Lung scatter	
SAD	100 cm	D	59.9 cGy	r_d equivalent square[a]	12.5 cm	r_d equivalent square[a]	12.5 cm
SSD_0	98.5 cm	SPD	106.0 cm	$S_c(r_c)$	1.012	$S_c(r_c)$	1.012
d_0	1.5 cm	d	15.2 cm	$S_p(r_d)\cdot\mathrm{TPR}(d,r_d)$ [a]	0.668	$S_p(r_d)\times\mathrm{TPR}(d,r_d)$ [a]	0.668
r_0	10×10 cm^2	d_{eff}	15.0 cm	OAR(d_{eff},x)	1.001	OAR(d_{eff},x)	1.001
D_0'	1 cGy/MU	r_c	22×13 cm^2			CF[a]	0.943
		x	0.9 cm				
		Missing scatter	lateral lung tissue				

[a]Calculated using Day's method. See text for details.

of Tables II and III and the remedial correction methods that could be applied in a range of different clinical situations. For each case, the MU or the dose to a selected point is compared to that from the primary TPS and the difference is analyzed according to the action level criteria in Tables II and III. All examples have been selected from actual clinical treatments, with primary MU calculated by a TPS with pencil beam or convolution-superposition algorithm using patient CT data. The data used in each MUV calculation are listed in a separate table and are classified as to the reference conditions used for the calculation, the description of the calculation point extracted from the plan description, and the associated dosimetric parameters used for the MUV calculation.

A variety of the methodologies discussed in Appendix A are used for the MUV to illustrate the different approaches that can be employed and the corresponding application of the criteria tables. Since some of these methods can be time-consuming to perform on a routine basis, the task group encourages the user to adopt an appropriate balance of effort, efficiency, and accuracy when implementing his/her MUV program. While the manual calculation formalism is used in these examples, the same approaches to resolving discrepancies arising from commercial MUV software can also be employed.

1. Whole brain case

The following example is a simple two-field whole brain case. The left lateral 6 MV photon beam was calculated. For this case, monitor units were calculated at a point in the isocenter plane that is marked with a dot in Fig. 3. The field flash was approximated as blocked field. The MUV calculation was determined from Eq. (A2),

$$\mathrm{MU} = \frac{D}{D_0' \cdot S_c(r_c) \cdot S_p(r_d) \cdot \mathrm{TPR}(d,r_d)[(\mathrm{SSD}_0 + d_0)/\mathrm{SPD}]^2 \cdot \mathrm{OAR}(d,x)}, \tag{B1}$$

where the factors for tray attenuation and wedge attenuation have been dropped as these accessories are not used.

The dosimetric parameters listed in Table VI yielded a MUV of 78.7 MU. The TPS calculated 79 MU. The agreement is within 1% and falls within the 3% action level for homogeneous conditions using different calculation algorithms and approximate patient geometry with minimal field shaping.

2. Wedged breast case

In the following tangent breast case, 6 MV photon beams were used, in conjunction with a 30° wedge. The treatment setting of 159 MU was obtained from a homogeneous TPS calculation for the medial field. The MUV calculation for the medial field was performed for a point positioned, as shown in Fig. 4(a). Three different calculation approaches were used in this example to deal with tangential beam flash and patient contour issues. Each of these approaches used the same equation, based on Eq. (A2),

$$\mathrm{MU} = \frac{D}{D_o' \cdot S_c(r_c) \cdot S_p(r_d) \cdot \mathrm{TPR}(d,r_d) \cdot [(\mathrm{SSD}_0 + d_0)/\mathrm{SPD}]^2 \cdot \mathrm{OAR}(d,x) \cdot \mathrm{WF}(d,r,x) \cdot \mathrm{CF}} \tag{B2}$$

for the plan parameters listed in Table VII.

a. Uniform cube phantom calculation

Calculating the MUV for a uniform cube phantom in which full scatter condition is assumed yields a MUV value of 149.0, which is −6.3% less than the TPS value and beyond the action level of 5% difference for different algorithms, uniform cube phantom approximation for an off-axis wedge distribution. In this situation, a more accurate estimate of patient geometry effects may be required to achieve an acceptable MUV.

b. Technique averaged patient geometry

The effect of missing scatter and contour change for standard breast treatment geometries has been investigated by several authors,[31,33,34] with each finding that the dose is reduced by a constant factor over a large range of phenotypes. Adopting a factor of CF=0.96 for standard breast treatment geometry from the literature[31] and using this in Eq. (B2), together with the factors appropriate for the uniform cube phantom approximations, yields a MUV value of 155 MU, which is within −2.5% of the TPS calculated MU setting and within the 4% action level for different algorithms, approximate patient geometry for an off-axis wedge distribution give in Table II.

c. Reduced field size scatter correction

The effect of missing tissue from tangential fields is estimated by treating the horizon of the breast seen in Fig. 4(b) as though it defines a block and so reduces the irradiated volume of tissue. This yields a value of 152.8 MU, which is within −3.9% of the 159 MU calculated by the TPS. This is

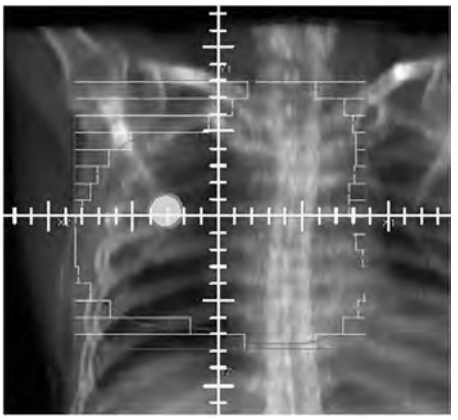


FIG. 7. BEV for AP lung treatment field described in Appendix B 5. The circle indicates the MUV calculation point.

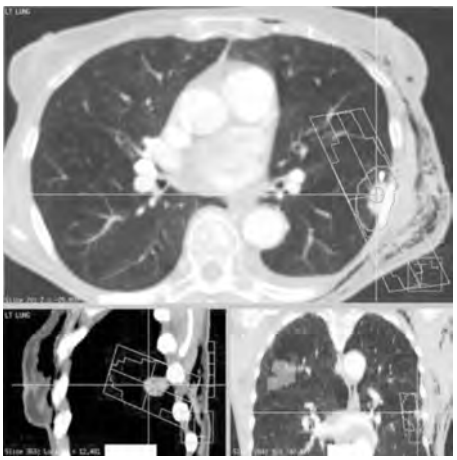


FIG. 8. Orthogonal views of MUV point within a small field lung lesion calculated in Appendices Secs. B 6 a and B 6 b. The isocenter was selected as the calculation point.

within the homogeneous action level for different algorithms, approximate patient geometry, and off-axis wedged fields in Table II.

3. Mantle case

The following example is a typical SAD mantle treatment using a 6 MV parallel-opposed pair. A heterogeneous TPS calculation for a point located in the AP field, as shown in Fig. 5, yielded a beam setting of 97 MU. The blocked area is defined by a mixture of MLC and poured blocks. The MU was verified using the equation,

$$\mathrm{MU} = \frac{D}{D_o' \cdot S_c(r_c) \cdot S_p(r_d) \cdot \mathrm{TPR}(d,r_d) \cdot [(\mathrm{SSD}_0 + d_0)/\mathrm{SPD}]^2 \cdot \mathrm{OAR}(d,x) \cdot \mathrm{TF}}. \tag{B3}$$

An independent Clarkson integration of the missing scatter, including blocks and tangential flash, gave an equivalent square of 19.4 cm at the point. Using the parameters in Table VIII, the MU calculated for the AP mantle field is 94.0 MU, which is 3.2% less than the TPS value of 97 MU. The agreement is within 3.5% and below the action level for different algorithm, approximate patient geometry, and large field listing in Table III. The MUV calculation could be improved by using an improved estimate of the blocked field scatter reduction which accounted for the near proximity of the lung field and would be expected to further increase the number of MU in the MUV calculation.

4. Modified mantle case

This example demonstrates the use of Day's method[66] to determine the effects of field shaping and reduced lung scatter when heterogeneity corrections are required. A modified mantle is treated with a parallel-opposed pair of 6 MV beams. The MUV calculation is presented for the posterior field for the calculation point shown by the circle in Fig. 6. The MUV was performed using the equation,

$$\mathrm{MU} = \frac{D}{D_o' \cdot S_c(r_c) \cdot S_p(r_d) \cdot \mathrm{TPR}(d,r_d) \cdot [(\mathrm{SSD}_0 + d_0)/\mathrm{SPD}]^2 \cdot \mathrm{OAR}(d,x) \cdot \mathrm{CF}}, \tag{B4}$$

with the parameters listed in Table IX.

a. Field shaping effects

Day's method was used to estimate the effect of shielding-reduced scatter in the term $S_p(r_d) \cdot \mathrm{TPR}(d,r_d)$ according to the equation,

TABLE X. Large field lung case.

Reference conditions		Treatment description		Dosimetric parameters: Homogeneous calc.		Dosimetric parameters: Radiological depth	
SAD	100 cm	MU	54 MU (45.17 open, 8.83 wedged)	r_d equivalent square	12.4 cm	r_d equivalent square	12.4 cm
SSD_0	100 cm	SPD	100 cm	$S_c(r_c)$ [a]	1.008	$S_c(r_c)$ [a]	1.008
d_0	1.5 cm	r_c	17×16 cm^2	$S_p(r_d)$	1.004	$S_p(r_d)$	1.004
r_0	10×10 cm^2	d	11.8 cm	$\mathrm{TPR}(d,r_d)$	0.749	$\mathrm{TPR}(d_{\mathrm{eff}},r_d)$	0.859
D_0'	1 cGy/MU	d_{eff}	7.7 cm	$\mathrm{OAR}(d,x)$	1.014	$\mathrm{OAR}(d_{\mathrm{eff}},x)$	1.015
		x	3.0 cm	$\mathrm{WF}(d,r,0)$	0.266	$\mathrm{WF}(d_{\mathrm{eff}},r,0)$	0.260
		Motorized 60° wedge	16.35% of MU	$\mathrm{WOAR}(d,x)$	0.987	$\mathrm{WOAR}(d_{\mathrm{eff}},x)$	0.987

[a]Based on MLC aperture equivalent square for a unit where the MLC replaces the upper jaw.

$$S_p(r_d)\cdot\mathrm{TPR}(d,r_d)=\tfrac{1}{2}(S_p(22.8\ \mathrm{cm})\cdot\mathrm{TPR}(15\ \mathrm{cm},22.8\ \mathrm{cm})-S_p(10.9\ \mathrm{cm})\cdot\mathrm{TPR}(15\ \mathrm{cm},10.9\ \mathrm{cm}))+S_p(8.4\ \mathrm{cm})\cdot\mathrm{TPR}(15\ \mathrm{cm},8.4\ \mathrm{cm}) \quad \text{(B5)}$$

based on calculating half the scatter dose from the symmetric rectangle centered on the calculation point (22×21 cm^2) less the central portion containing the narrowed shielding (22×7 cm^2), with the last term accounting for the reduced shielding area approximately centered on the calculation point (10×7 cm^2). The equivalent square field sizes were projected to their size at the SPD. Using the data from the clinic's dosimetry tables yields a value for $S_p(r_d)\cdot\mathrm{TPR}(d,r_d)=0.668$, corresponding to an equivalent square of approximately 12.5 cm. Using these values in Eq. (B4) generates a MUV value of 99.6 MU, which is 5.3% below the primary MU calculation, and beyond the action level of 3.5% suggested in Table III for this type of calculation.

b. Field shaping and lung scatter effects

A further correction factor to account for reduction in lateral scatter due to reduced tissue density is obtained by treating the patient geometry as though the calculation point was embedded in unit density mediastinal tissue, surrounded laterally by reduced density lung tissue. In this case, the mediastinal tissue is approximated as a BEV region with dimensions of 5×12.5 cm^2, embedded in a 12.5×12.5 cm^2 (equivalent square) region of lung tissue. Equation (A7) then takes the form,

$$\mathrm{CF}=\frac{S_p(7.1\ \mathrm{cm})\cdot\mathrm{TPR}(15\ \mathrm{cm},7.1\ \mathrm{cm})+\rho_{\mathrm{rel}}\cdot(S_p(12.5\ \mathrm{cm})\cdot\mathrm{TPR}(15\ \mathrm{cm},12.5\ \mathrm{cm})-S_p(7.1\ \mathrm{cm})\cdot\mathrm{TPR}(15\ \mathrm{cm},7.1\ \mathrm{cm}))}{S_p(12.5\ \mathrm{cm})\cdot\mathrm{TPR}(15\ \mathrm{cm},12.5\ \mathrm{cm})}$$
$$=\frac{\rho_{\mathrm{rel}}\cdot S_p(12.5\ \mathrm{cm})\cdot\mathrm{TPR}(15\ \mathrm{cm},12.5\ \mathrm{cm})+(1-\rho_{\mathrm{rel}})\cdot S_p(7.1\ \mathrm{cm})\cdot\mathrm{TPR}(15\ \mathrm{cm},7.1\ \mathrm{cm})}{S_p(12.5\ \mathrm{cm})\cdot\mathrm{TPR}(15\ \mathrm{cm},12.5\ \mathrm{cm})}. \quad \text{(B6)}$$

TABLE XI. Initial calculation point position in B6 small field lung case.

Reference conditions		Treatment description		Dosimetric parameters: Blocked equivalent square		Dosimetric parameters: O'Connor's theorem	
SAD	100 cm	D	351.8 cGy	r_c equivalent square	3.5 cm	r_c equivalent square	3.5 cm
SSD_0	100 cm	SPD	100 cm	r_d equivalent square	3.5 cm	r_d equivalent square[a]	1.1 cm
d_0	1.6 cm	d	13.8 cm	$S_c(r_c)$	0.946	$S_c(r_c)$	0.946
r_0	10×10 cm^2	d_{eff}	4.2 cm	$S_p(r_d)$	0.971	$S_p(r_d)$	0.951
D_0'	1 cGy/MU	x	0.0 cm	$\mathrm{TPR}(d_{\mathrm{eff}},r_d)$	0.937	$\mathrm{TPR}(d_{\mathrm{eff}},r_d)$	0.927
				$\mathrm{OAR}(d_{\mathrm{eff}},x)$	1.000	$\mathrm{OAR}(d_{\mathrm{eff}},x)$	1.000

[a]Scaled according to O'Connor's theorem. See text for details.

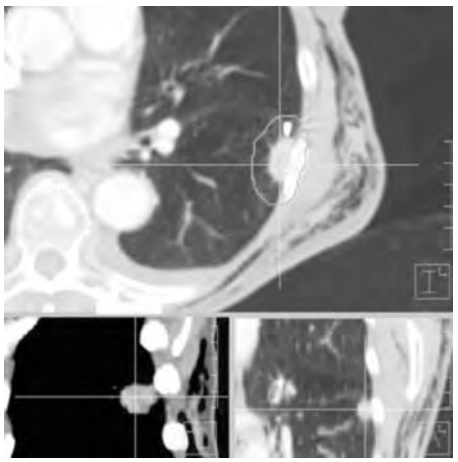


FIG. 9. Orthogonal views of alternate MUV calculation point placed within the center of the normal density tissue within the target volume, discussed in Appendix B 6 c. The selected point is placed away from tissue interfaces as discussed in Sec. V A.

Using a relative density $\rho_{rel}=0.25$ for lung and the clinical values of $S_p(7.1\text{ cm})=0.990$ and $\text{TPR}(15\text{ cm},7.1\text{ cm})=0.624$ in Eq. (B6), a value of CF=0.943 is obtained, and the resultant MUV value of 105.6 MU falls within +0.6% of the TPS derived value, well within the 3.5% action level of Table III.

5. Large field lung case

In this example, the MUV takes the form of an independent prediction of the dose to a point rather than a determination of MU. The target in the lung is treated using 6 MV photon beams in a four-field technique. The calculation point is 3 cm off-axis at SPD=SAD=100 cm (blue circle in Fig. 7). The MUV calculation is provided for the AP beam, which uses a 60° motorized wedge for 16.35% of the MU. The primary MU calculation used a convolution-superposition algorithm with heterogeneity corrections. The MUV calculation was done with and without heterogeneity corrections to illustrate the need to use corrections in the verification whenever it is used in the primary calculation. Equation (A1) is used

$$D=\text{MU}\cdot D_0'\cdot S_c(r_c)\cdot S_p(r_d)\cdot\text{TPR}(d,r_d)\cdot[(\text{SSD}_0+d_0)/\text{SPD}]^2\cdot\text{OAR}(d,x)\cdot\text{TF}\cdot\text{WF}(d,r,x) \quad \text{(B7)}$$

together with parameters listed in Table X.

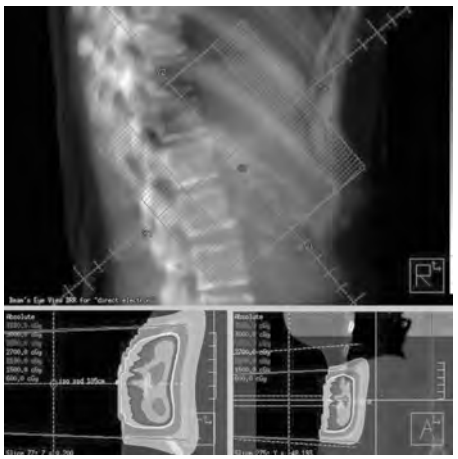


FIG. 10. BEV for a single direct electron field calculation of Appendix B7, with the calculation point indicated by the circle.

TABLE XII. Adjusted calculation point position in B6 small field lung case.

Reference conditions		Treatment description		Dosimetric parameters	
SAD	100 cm	D	354.4 cGy	r_c equivalent square	3.5 cm
SSD_0	100 cm	SPD	100 cm	r_d equivalent square	1.1 cm
d_0	1.6 cm	d	13.8 cm	$S_c(r_c)$	0.946
r_0	10×10 cm^2	d_{eff}	4.6 cm	$S_p(r_d)$	0.951
D_0'	1 cGy/MU	x	0.4 cm	$\text{TPR}(d_{\text{eff}},r_d)$	0.902
				$\text{OAR}(d_{\text{eff}},x)$	1.001

a. Homogeneous calculation

Using the factors for a homogeneous dose calculation, a dose of 37.6 cGy from the field is obtained, which differs from the TPS calculated dose of 41.9 cGy by 10.3%, which is beyond the 5% action level of Table II for different algorithms, uniform cube phantom with wedge.

Radiological depth correction

Using the radiological depth reported by the TPS to derive the dosimetric parameters yields a dose of 43.1 cGy, which differs by 2.9% from the TPS predicted dose, and is within the large field action level of 3.5% for approximate patient geometry in Table III.

6. Small field lung case

The following example is a ten-field lung case using 6 MV photon beams. The RAO field, shown in Fig. 8, requires 443 MU as determined by a commercial TPS using a convolution-superposition algorithm performing a heterogeneous calculation. Adapting Eq. (A2) by dropping terms for accessories not used in this treatment, the verification MU was calculated according to

TABLE XIII. Direct electron case.

Reference conditions		Treatment description		Dosimetric parameters	
SAD	100 cm	$D(d)$	300 cGy	$S_e(r_a,6\text{ cm}\times6\text{ cm})$	0.995
SSD_0	100 cm	SSD	105 cm	$S_e(r_a,10\text{ cm}\times10\text{ cm})$	1.005
d_0	3.0 cm	d	3.6 cm	$S_e(r_a,6\text{ cm}\times10\text{ cm})$ [a]	1.000
r_0	10×10 cm^2	PDD	97%	g	5 cm
D_0'	1 cGy/MU	r_a	10×10 cm^2	SSD_{eff}	86 cm
		r	6×10 cm^2		

[a]Calculated from the square aperture S_e values using Eq. (A10).

$$\mathrm{MU} = \frac{D}{D_o' \cdot S_c(r_c) \cdot S_p(r_d) \cdot \mathrm{TPR}(d,r_d) \cdot [(\mathrm{SSD}_0 + d_0)/\mathrm{SPD}]^2 \cdot \mathrm{OAR}}. \quad \text{(B8)}$$

In this case, two different methods were used to estimate the reduction in scatter dose from field blocking and patient geometry. The use of a different calculation point to achieve agreement within tolerance was employed in a third calculation.

a. Standard blocked equivalent square

The dosimetric effects of scattered radiation were characterized only by the blocked field area with a radiological depth used for density corrections. Using the values found in Table XI in Eq. (B8), the MUV calculation for the RAO field was 396 MU, which is 12% less than the TPS calculated value of 443 MU, and outside the action level in Table III for different calculation methods and small fields with inhomogeneity corrections.

b. O'Connor's theorem for blocked equivalent square

A more detailed treatment of the scattering conditions experienced by the calculation point within the patient was obtained based on O'Connor's theorem,[52] which holds that the equivalent square field characterizing the reduced scatter, $\mathrm{EqSq}_{\mathrm{eff}}$, can be approximated by the equation, $\mathrm{EqSq}_{\mathrm{eff}} = \mathrm{EqSq}_{\mathrm{blocked}}(d_{\mathrm{eff}}/d)$. An equivalent square of 1.1 cm is obtained, generating the parameters listed in Table XI under the heading of O'Connor's theorem. Using these values, the MU calculated for the RAO field is 409 MU, which is within 8% of the TPS value of 443 MU. The MUV is improved from the simpler standard block treatment, but is still outside the action level for different calculation methods and small fields.

c. New calculation point

An examination of the point placement in Fig. 8 raised concerns that the calculation point may be positioned too close to an interface and may be subject to rebuild-up and lateral electron disequilibrium. To reduce such effects, the calculation point was moved to the center of the tumor, as shown in Fig. 9, and O'Connor's theorem was applied to obtain a MUV calculation. The parameters for this calculation are given in Table XII, for which the MUV was determined to be 422.8 MU. This MUV is within −4.6% of the 443 MU calculated by the TPS and within the 5% action level for different calculation methods and small fields given in Table III.

7. Direct electron case

In this example, an en face 12 MeV electron beam was used to treat a rib (see Fig. 10). A rectangular aperture of 6 $\times 10$ cm^2 in a 10×10 cm^2 applicator was used at 105 cm SSD. The TPS calculated dose distribution was normalized to a nominal depth of 3 cm and prescribed to deliver 300 cGy to the 97% isodose line.

The output setting was determined according to the equation,

$$\mathrm{MU} = \frac{D}{D_0' \cdot S_e(r_a,r)} \cdot \left(\frac{\mathrm{SSD}_{\mathrm{eff}} + d_0 + g}{\mathrm{SSD}_{\mathrm{eff}} + d_0} \right)^2. \quad \text{(B9)}$$

The MUV was calculated using the dose D, derived from the prescription of 300 cGy/97% = 309.3 cGy, and the parameters listed in Table XIII. Using these parameters, a value of 345 MU was obtained, which is within 0.9% of the TPS calculated value of 348, and within the action level of 3% in Table II for different algorithms, uniform cube phantom, and modest field shaping.

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