

This is sort of a continuation of the previous refresher course; this is a report of Task Group 71 which is a task group for monitor unit calculations for photons and electrons. I'm John Gibbons and Don Roback with me from Maplewood Cancer Center in Minneapolis, Minnesota. And here's an outline for today's presentation, we'll begin by talking about the task group itself, its formation and charge and a little bit of an overview on calculations in general, we'll delve into photons calculations, then we'll move into electron calculations and talk a little bit about quality assurance and then go into some conclusions. As you all are probably aware, there is no current protocol within the AAPM for doing monitor unit calculations and recognizing this need three years ago, Task Group 71 of the Radiation Therapy Committee was created to create a standard formalism for doing these calculations. The membership consists primarily of myself and several,

Don, and several others, most of these people participated in a symposium that we did in the Southeast chapter in 1999, after which we had decided to propose this task group to the RTC. And the task group charge consists of this, actually multi-faceted, begins by emphasizing the importance of using a unified methodology. Right now in this country there are a number of different formalisms in use for calculating monitor units. That makes it difficult when people move between clinics to compare tables, to understand for dosimetrists what different parameters mean, whether or not their magnitudes are accurate and so forth. And as a result of that, we will recommend a use of a consistent terminology, which is a lot more difficult than it sounds, getting a large group of people to agree on such, recommend particular algorithm associated with those dosimetric variables, recommend measurement and/or calculation result methods for determining

these dosimetric variables, recommend quality assurance tests for not only existing computer programs which are used to calculate monitor units but also on hand tables and hand algorithms that are used within your clinic and then provide example calculations for common clinical situations. Again as I mentioned, it was formulated in 2001, at our first meeting three years ago, at this meeting, first face-to-face meeting and as typical for task groups that are proposed, this one had a two year sunset date and it was extended last year. A draft report was submitted to the Radiation Therapy Committee this year and the final report is due by the end of this year so hopefully by this time next year we'll have something final out. And this is an outline of the report, begins with an introduction to the importance of monitor unit calculations and some general concepts and then delves right into the nomenclature similar to TG-51, we begin by defining the

variables and defining the symbols that are associated with each of the dosimetric variables and we're gonna talk a little bit about those. This actually took quite a bit of time to come to an agreement between not only the members of our committee, but also the members of the Radiation Therapy Committee as well. Next we'll go right into the calculation formalism and begin by telling you what the equations are and then perhaps just as important, how to decide what the input parameters for these equations are, what is the depth that's appropriate to use, what is the field size that you should use when you're doing a calculation for particular clinical setup. After all of this is done, of course, it's necessary to create the tables for the, the data that you used for generating those so

section four is probably the longest section of the report where we go into how to measure all the quantities, all the output factors, percent depth doses, etc. Important

sections on interfacing to treatment planning systems and quality assurance follow and then, then with examples. So as an introduction, it's important to recognize what is the, the objective that we're trying to achieve here, and the ICRU-24 recommends that overall dosimetric systems should be capable of providing accuracy within plus or minus five percent. Now that's for the entire prediction of dose and calculation of monitor units is only one step of that so it's clearly necessary to have better accuracy within monitor unit calculations. It's important, however, to remember that there is a difference between absolute and relative dosimetry. Absolute dosimetry, which would be the determination of absolute dose per monitor unit at a particular point, is really governed by several international and national protocols, TG-51 in this country, for determining the dose per monitor unit. That has its own intrinsic error which if, of course, would

be combined with the error associated with monitor unit calculations, which we'll talk a little bit about the distinctions between these two types of calculations later on. It's also very important to be consistent with your treatment plan as far as beam weighting, whether or not the treatment plan has already considered variables that you don't want to double correct for within the calculation itself and finally it's important to have a quality assurance program in particular these days, where calculations have gotten very complex. So as I mentioned there is a difference between absolute and relative dosimetry and within the protocol we very clearly try to distinguish between reference conditions and normalization conditions and by reference conditions we're really referring to those conditions that define the conditions for measurements of absolute beam output or absolute dose per monitor unit at a particular point. For example, in TG-51 the quantity

d_{ref} , is the reference depth and that's set for ten centimeters. That's to be distinguished from normalization conditions which for monitor unit calculations are important in this report and those are the conditions in which all the relative dosimetry functions are normalized, that is they're set to unity and in this report as you'll see in a moment, we call our normalization condition the quantity d_0 . In most clinics, the normalization condition in this country is set to the maximum depth of maximum dose or d_m . We're going to recommend a greater depth as we'll see later on. But nevertheless there's a difference between the normalization conditions and the reference condition as exemplified here. In this particular case we have a diagram of a beam incident on a phantom, the normalization depth d_0 may not necessarily as well as other parameters - SSD and field size, may be

different from that for the reference conditions and it's important to distinguish between those two. For photon beams, the normalization depth d_0 may be different from d_{ref} and it may also be different from d_m : You may have three different depths that are associated with your calculations, it's important to keep that in mind as we proceed. So now let's talk a little bit specifically about photon calculations, I want to talk about nomenclature and I'm going to present some standard symbolism that we've decided on in this report and finally come to an agreement with the Radiation Therapy Committee and in coming

to this symbolism for the various parameters within the protocol, we sort of had the three laws of nomenclature and I don't know if, if any of you have seen I Robot, but this is similar to the I Robot three laws, so anyway (pin drops). I thought it was funnier than, than.... But the first law of

nomenclature is to use commonly understood symbols, one of the things is in 1997, ESTRO published a report for formalism for calculating monitor units, very consistent, very accurate, but probably not used certainly not in this country and not greatly used in Europe and one of the difficulties was that much of the symbolism was changed from that which is commonly used within the literature, certainly within this country. So we wanted to make sure that when people opened up the protocol of the task group that it didn't become a whole lot of new symbols that people weren't used to so if, if possible, we wanna use commonly used understood symbols. The second principle that we used was to maintain consistency with other task group reports, Task Group 74 was one which just preceded us, also Task Group 70, which is rewriting the electron beam, Task Group 25 report also considers electron beam output and so forth and we want to

make sure we're consistent with those as well. And we'd like to do such things as avoid multiple letter subscripts in our variables, that was some of the principles that were used in the ESTRO report, but if it conflicts with using commonly understood symbols, then we didn't want to do that. So, with that in mind, here is a series of nomenclature that's in the current draft, the dose rate at a normalization, first I've separated this into first constants, these are variables that are non-variables, D_0 , is the dose rate at the normalization point and in many clinics these days that's set to one cGy per monitor unit for a 10 by 10 field size, at 100 SSD and a depth of d_m , but that doesn't necessarily have to be that way, within the formalism within the protocol. The reference depth of, actually sorry, the normalization depth, is set to d_0 , the normalization field size is r_0 , (all of the field size parameters are

labeled as r) The source to axis distance or source to isocenter center distance is SAD and SSD_0 represents the source to surface distance under normalization conditions. These are all constants, which do not change. The independent variables which are important, and we will talk about how to determine these for the calculations are listed here, all the depth parameters are given by the symbol d , in general the depth to the point of calculation is d , there's some specific depths, for example, d_{eff} would be the effective or radiological depth accounting for the heterogeneities, if you have used those within the calculation within your clinic. The depth of maximum dose, which is, we'll see later on, is a function of field size, it's given the quantity d_m . The field size of the surface is labeled r , r_d and r_{c} represent the field size of depth d , and the field size or the equivalent field size

defined by the collimator r_c . All of these quantities are consistent with that given in Khan's textbook. Additionally, other independent variables include source to point distance, or source to the calculation point distance for doing isocentric calculations to points other than at the isocenter, SSD for the source to surface distance, and finally, the off axis distance; the quantity x is used, but it's emphasized that in this capacity x is the radial distance from the central axis in the plane of calculation. So most of these symbols

are fairly common, most people wouldn't be intimidated by seeing these nomenclature and that's the idea is to make sure that the quantities themselves and the algorithm is consistent but not daunting for those people attempting to put it in place. Finally the dependent variables, those functions which we'll attempt to determine the dose to the calculation point is D . Off axis ratio we use OAR. Percentage depth dose normalized

to d_m , that which, which, what, what were commonly used is PDD, but in this protocol we use what's called a normalized percentage depth dose where the percent depth dose is normalized to the depth of normalization. Since we recommend later on that the depth of normalization not be at d_m , this is a quantity that's probably unfamiliar to most, but this given by the symbol PDDN. The output factor for photons, the total output factor is S_{cp} and it's assumed to be separable into two components: the in-air output ratio or sometimes known as the collimator scatter factor, S_{c} and the phantom scatter factor S_p represent those components which are due to scatter within the phantom or to scatter within the rest of the collimator, the flattened filter, etc. Tissue phantom ratio is TPR, tissue maximum ratio as defined within Khan's textbook is really a special subset of

the tissue phantom ratio where the depth is at d_m , it's not used explicitly within the protocol. Tray factor and wedge factor are also given by these fairly common nomenclatures. So, that represents the nomenclature and with that we'll go straight into how we calculate monitor units within the protocol but before we do, I want to have a little bit of a discussion about the choice of depth of normalization and it's important that when you choose one within your clinic and again the protocol doesn't necessitate the choice of any particular one, but there are certain restrictions about what choices can be made. But it's important to remember that all quantities are supposed to be determined at this depth so if you choose a normalization depth of d_m , the wedge factors, the output factors, all of these things should be determined at this depth. It's required within the protocol, that the normalization depth is greater than the

maximum depth of maximum dose, the maximum d_m , and the idea behind this is that we would like to at least be on the range of electron contamination. And how do we determine the maximum depth of maximum dose? Well that is simply an extrapolated maximum d_m for the smallest field size and actually for the largest SSD as well, so this represents how to determine what d_m is appropriate for your users beam. For example if you've got a new energy beam, you're scanning it and you get multiple depth doses, then you want to extrapolate those depth doses back to the smallest field size actually going back to zero to determine what your d_m is. TG-71 however, recommends that we will use a normalization depth of 10 centimeters. Why ten centimeters, there's a number of reasons, I've listed three of them here. First is, there's a better consistency with our calibration protocol, implicit

within the conversion from TG-51 to the normalization point, will be some type of depth dose or tissue phantom ratio which inherently adds another step in the process and leads to a potential error so as long as we use a normalization depth equal to that of the reference depth, when we do our calibration we are effectively determining the dose of the normalization point and that step is eliminated. The second argument is given in the

ESTRO protocol which also recommends a depth of ten centimeters and that is that in reality, the percent depth dose of d_m is inaccurate for photon beams. As you know, we typically use the percent depth ionization for the percent depth dose in photon beams but that's assuming that the stopping power ratios are virtually unchanged when we convert from ionization to dose at deeper depths. But when we get to d_m , we know that the spectrum is much different because of contaminant

electrons and that argument is not truly valid. So there is some inaccuracies associated with determining the percent depth dose of d_m . Finally, there are, there is electron contamination even in spite of the fact that we try to create the maximum depth of maximum dose, we still create problems, particularly for high energy photon beams and we'll look at some of those problems a little bit later on. However, with this caveat, it's important to know that Task Group 71 formalism is valid for a normalization depth equal to the maximum depth of maximum dose and can be used if that is the choice of normalization depth within your clinic. So, monitor unit calculations as opposed to absolute dosimetry really represent a relative calculation and we have to go through the steps of relating the dose of an arbitrary point to the dose under normalization conditions which we're all familiar with or which with we've established within our

clinic. So here's a basic diagram of the derivation of the algorithm for isocentric calculations but as you can see there's multiple steps from an arbitrary point which may be a particular distance x off-axis, particular field size of r_d . It may have blocks, it may have a wedge in place and all of these parameters are then related to parameters until we eventually go from in this case a) all the way to g), which is our normalization conditions and this is how we determine the process of getting the dose from the normalization point to the dose of any arbitrary calculation point. When we go through this process, if you want to calculate the dose to the isocenter we get this equation here which most of you are familiar with, the monitor units for calculation to the isocenter, the dose prescribed to the point of calculation which is the isocenter divided by the dose rate at the normalization point times the collimator scatter factor and in this particular case I

put the collimator field size but we're going to talk a little about that in a moment. The phantom scatter factor and the tissue phantom ratio are dependent on the field size at depth and, of course, the tissue phantom ratio has a depth dependence as well. The wedge factor may or may not have a strong depth or field size dependence, but in this case we've put as default that we do have a depth and field size dependence, most clinics do not currently do that according to the RPC, but we recommend within the protocol that that's done. Any additional attenuation factors are contained here and finally an inverse square term, if the point, the calibration point, the point where the normalization is, is SSD_0 plus d_0 , the normalization point is not at the isocenter. Now if we have a calculation to some other point that's not necessarily at the isocenter, we can still do this calculation isocentrically and here is the generic or the calculation

that's asso-, for any arbitrary point, we've added the fact that the again, there's an inverse square term, if that's not at the point of calculation. Instead, however, of using the source to axis distance, we use the source to point distance which represents the distance to the

point of calculation. There is an off axis ratio that's also in place if we happen not to be on the central axis as well. The protocol very strongly recommends that all calculations are done isocentrically, it makes the process much easier. However, we do bring up SSD or percent depth dose calculations. This is the formalism for doing this type of calculation but it's a little bit more complex from the previous one and primarily that's because of the field sizes that go in place and this, this particular case, the monitor units of the dose divided by the dose rate at the normalization point, again there's collimator scatter which in this case is dependent on the collimator jaws and we'll

talk again about that in a moment. But then the field sizes that are used for the rest of the calculation present a bit of a problem, the percent depth dose is defined by the field size at the surface, as are other parameters but also critical for us in addition to the collimator field size and field size at the surface, is the field size at the reference depth. Now, in normal circumstances, and if you go through, for example, that published within Khan, this is just listed as the field size of the surface but if you do a rigorous derivation from tissue phantom ratio formalism to a percent depth dose one, then actually this is the field size of the reference depth. If you choose the reference depth near d_m , then it turns out that there's very little difference because there's not a lot of divergence from the surface to the depth of maximum dose. However, if you have a reference depth of ten centimeters, there's more divergence. Not only that at 10

centimeters, the phantom scatter factor is a much steeper dependent, has a much steeper dependence on field size. So those two facts make it impossible to ignore that if you want to be within a percent. How do we determine the field sizes that go into those three equations for determining monitor units and, and we're going to talk a little bit about that. As an overview if you have a rectangular fields, and you have a table of square values, you need to use the method of equivalent squares to determine what is the equivalent square field for the effective field that's actually incident on the patient. For rectangular fields, of course, there's an equivalent square approximation for A over P and if you have highly irregular ones, you can sometimes look up equivalent square tables, either of these are appropriate within the protocol. If you have something that's highly irregular, you may require Clarkson integration. All of these types of

approaches are valid for any type of phantom scatter, scatter within the phantom. However, for any type of scatter within the head which may change based on geometry and so forth, these relationships should be verified experimentally. Now I want to talk about determination of field size for the specific quantities, first for S_c , and I'm gonna talk a little bit about this, I know for those of you that were in the earlier session, you probably heard a lot more. But within the protocol for open or block beams, the primary method, mechanism for determining the equivalent square is to use a collimator field size, that is that field size divided the isocenter, given by the symbol r_c . Now there are more accurate methods that are available, for example, the point's eye view model, that's given in the appendix of the protocol, and we're going to look at that in just a moment, and that may be required if you've got, for example, a rectangular

field with a very large aspect ratio 5 by 30 or more, or if you have a highly any type of highly irregular field defined by blocks or MLC's. And then, when, look at the point's eye view model and to understand why this becomes important. Again in the way, the mechanism to do that is to look at where the source of head scatter comes from within Sc . There's a couple of sources to head scatter here that we'll deal with, there's back scatter to the monitor chamber as you can see from this arrow, photon beams coming down, electrons back scattering into the chamber cutting off the dose rate depending on how far the jaw's extended. Head scatter itself can come from two sources that are listed here, primarily the adjustable collimators themselves off the edges of the corners of the blocks where transmission through the very, the edges, or from the flattening filter and studies have shown that the flattened filter is the dominant mechanism for

where the head scatter comes to the point of calculation. So now if we're interested in the dose of the point here and the flattening filter is positioned here, then what happens is we're really interested in how much scatter can make it to the point of interest here and not be blocked by these jaws and notice that it's a little bit of a different scenario than these jaws being opened to see how wide the field is when it gets to the isocenter. Now we're interested in how much of this scatter dose can be seen by this point down here and this is the so called point's eye view or detector's eye view model of head scatter that determines what is the magnitude of the head scatter that's going to come down to this point. So if we consider, for example, a ten by ten field size and we've taken the lower jaws and rotated them around so that they're in the same plane so we can visualize them so they're, they're basically on the same divergence plane, you can see

that from this point, there's a, there's a big difference in the field that it sees in the flattening filter. The upper jaws restrict the view of this point of the flattening filter, a lot more than the lower jaws do. So for a ten by ten field or, which is shown by this square here, if we look from the point's eye view up at the flattening filter based on geometry and this case it's roughly the geometry of a Varian machine, you see that the field size that's seen on the flattening filter is this field size here, it's rectangular and of course it's much smaller, but the, the idea is that it's not the same size because the upper and lower jaws restrict differently the view of the flattening filter which then again restricts the amount of, of collimator scatter that's going to make it to this point. Also shown down here are blocks which would restrict the beam incident on the patient somewhat but as you can see from their placement, they have very little restriction of the flattening filter scatter and

thus the block field size has little effect on the collimator scatter. So, I'm going to talk a little bit about the collimator exchange effect, the collimator exchange effect is defined as the value for the in air output ratio, or Sc , when the parameters A and B are the two field sizes for example, the upper and lower jaw field sizes do not equal, does not equal the collimator scatter when the low jaw field sizes are reversed so in other words, you have a collimator scatter associated with 5 by 40 versus 40 by 5 by swapping the jaws and settings and you get different output and that's demonstrated for both open and wedge fields, and in most, most cases, and this is important, it's typically less than two percent. And here it is shown graphically, here this is a ratio of, for collimator scatter factors for

square field sizes which are shown by the black squares here, and in this case the square fields are varied from 5 by 5 to 40 by 40 and also what's done,

is we've set one jaw to 40 centimeters and varied the lower jaw, varied the other jaw as a function of field size and looked at the collimator, the resultant collimator scatter relative to a 10 by 10 and you can see that when you vary the upper jaws or vary the lower jaws, you get a difference in the collimator scatter depending on which jaw is varied. From the diagram on the point's eye view, of course this makes sense because the upper jaws restrict the view of the flattening filter more. From the point's eye view, if we argue that what's important rather is how much of the flattening filter we view, then if we re-plot these data, based on their flattening filter field size rather than the equivalent field size at isocenter, what you see is that all the data line up very well, certainly within, well within a percent which is the difference between these two horizontal, with the horizontal lines on the graph. So you can do a much better job if you use the equivalent field size

at the flattening filter. That's a fairly difficult thing to do and many computer programs will do this, but and certainly within the hand calculation it's difficult. The good use is that if we do use the equivalent field size at isocenter, the difference between the data that we see even in a maximum 5 by 40 for this machine is barely over two percent. So in most cases, using the equivalent field size at the isocenter for S_c , is sufficient. What happens if you have other apertures other than blocks or blocks use, you typically don't avoid it, MLC's are a different story, again under the point's eye view model and that's, and actually only those apertures that are closer to the flattening filter are going to influence S_c , thus the field size argument, that is, what, what do we use for the field size within that dosimetric parameter will depend on the MLC location. So the upper collimators and their three, so their MLC's can either be in upper or lower

jaw, collimator replacement or tertiary MLC, and studies have shown that, here's a diagram of these three replacements, upper jaw replacement typically offered by Lecta, lower jaw replacement by Siemens's or tertiary by Varian and a number of others. These three have different arguments for what you should use for the, for the S_c . For the upper and lower jaws studies by Jatinder Palta and Indra Das have found that S_c is actually better described by using the MLC field rather than the collimator field to define what the collimator scatter is. But when the jaws are placed, when the MLC's are placed below the lower jaws, Eric Klein found that the S_c was better described by the collimator jaws rather than the field size defined by the tertiary collimator. Now there are other parameters rather S_c , and we have to determine what the field size is for used, for those, and in those

cases they're more effected by the amount of phantom scatter rather than the scatter within the flattening filter or within the air or the collimator jaws, etc., and we define the effective field size as the equivalent square of the field size incident on the phantom, and this field size of course would be reduced by their custom blocking or MLC's or any fall off or missing tissue. For S_p within their algorithm, just to remind you that the effective field size within the isocentric formalism is the field size at depth within the SSD formalism, it's at the normalization depth for the tissue phantom ratios and wedge factors

incident use the effective field size at depth and for percent depth dose you use the effective field size at the surface. Now we've talked about field size for use in the calculations for determination of depth is simply the depth of the calculation point unless you're using heterogeneity corrections in which case you use the

effective or radiological depth, d_{eff} . It's not universally used, so therefore you have, it's important that your physicians become aware of whether or not you're going to use heterogeneity corrections, whether or not they have, are used to using heterogeneity corrections or if you're switching to a new planning system, it's important to let them know what kind of changes are going to be introduced if you've changed from one type of heterogeneity algorithm to another. There's a couple of possible methods for manual calculations that are done simply in the, in house, probably the simple-, these two path-linked models, the simplest is the ratio TAR where we look at basically the effective density of effective depth to the, to the point of calculation and if we wanna get a little bit more accurate we can use the Power Law TAR method or Batho method but that's, that's very difficult unless you have only two, one or two heter-, one

heterogeneity to do manually. Okay, so that's the use of, of the calc-, of those calculation parameters and how to determine what goes into them, next we want to talk about how to measure or determine the dosimetric parameters. And first we'll talk about output factors and I know you've heard a lot of it, for those of you who were here earlier, but the total output factor for S_{cp} is measured in phantom at the, at the normalization depth, it's important to separate collum and phantom scatter and that's emphasized by, within the protocol, and S_p is usually determined indirectly. In other words, the measurement of the total output factors divided by the measurement of the collimator or in-air output ratio to determine S_p . S_c is measured in air at the normalization depth and I can see that I've used reference myself several times, I didn't mean to, traditionally it's measured with a build-

up cap but if we use a larger depth of normalization, it's gonna require the use of mini-phantoms and you should make sure you avoid scatter from surrounding structures when you do this measurement, support stands, floor, wall, etc. Now, here's just a couple of graphs of output factors that are measured just to show you the differences when you convert or if you convert from dm-formalism to a depth of 10, as you can see the total output factor varies greatly for this, this is for a 6X beam, the green represents the output that you would measure at a depth, if you're normalizing to a depth of dm. The red is a normalization a depth of 10 and there's quite a substantial difference in the output as you go with a larger field size at a depth of 10 because you have more scatter contributing when you get to the lower depths. These lines represent two percent variations so they give you an idea of how significant, how

important these are. If we look at the individual components, the phantom scatter shows quite a bit of variation as well, determined by indirectly as the previous equation is shown, but the collimator scatter actually doesn't show, shows very little difference between 6 and a depth of 1.5 and depth of 10, and that's, this is from our clinic, it's very consistent with others. This isn't always true, for example, at high energies, at 24 MV,

this report from Frye et. al., at University of Wisconsin show that they saw substantial differences for high energy beams depending on the amount of overhead and overlaying material that they were used within the many phantom and for example here you see on the order of almost five percent difference in the measurement of S_c , and the, the question is why, why are you getting that much difference, the conclusion of the article was that it's important to normalize and measure everything at the

normalization depth, but there, it does create problems from dm-formalism, for example if we look at this simple diagram of determining the dose per monitor units or the number of dose per monitor units for two simple situations. For example here we have a collimator field size defined by some, some jaw position here and we have blocks here we're measuring at certain field size defined by the blocks and certain depth and, and then we simply open the collimator jaws to increase the collimator size leaving everything else the same, the ratio of the dose rates of these two quantities should be strictly determined by the field size of the collimators, the only change is that, and that only influences S_c at least according to the algorithm proposed and that's shown here, the ratio of dose rates for these two points is ratio of S_c 's. However, this is irrespective of the normalization depth, that doesn't come into

play in this equation at all so since we have S_c 's which vary by almost five percent depending on whether we use a mini phantom or build-up cap, the question is which one should we use and the explanation is found in that when you use dm-formalism for these high energy photon beams, some of the other quantities actually have a field collimator field size dependence such as S_p and TMR and these are for example 18X TMR's which are measured with collimator field size of 20 by 20, blocked to a 6 by 6 and a collimator field of 6 by 6 blocked to a 6 by 6 and we see difference between these two graphs, this is about ten percent intervals so it's quite a, it's about a three or four percent difference so on the unique situations where you have a heavily blocked field for high energy photon beams, you can have some problems with, if you do use a dm type of algorithm. For wedge factors, we have

two types of wedges as you, as you know, there's either internal wedges or external wedges depending on whether they're mounted above or below the collimator jaws and they've been extensively studied for their field size dependence but in some, the RPC has found that the wedge factor should have a field size dependence if the wedge factor's less than .65 and that's shown in this data here which shows a smattering of data from a number of clinics that shows the ratio of the wedge factor for 20 by 20 to a 10 by 10 and you can see that, that ratio is more than two percent if the wedge factor gets small. It also has a depth dependence, again with the same paper the RPC found that it did have a depth dependence if the wedge factor was smaller, less than 10 MV although traditionally many people have shown that the depth dependence is not too great if you don't go to depths less than 10 centimeters. And this is again that same report for

wedge factors but the recommendations of the protocol will be to have tables of depth and field size dependence. For filterless wedges, dynamic wedges or wedge factors, there's a number of different algorithm for determining what those wedge factors are for

EDW for virtual wedges of course at the central axis is very close to unity for all wedge angles and field sizes but if you go off axis, you actually have an exponential off-axis relationship. These are data from Eric Klein's group that show that they can get very good agreement using an analytic approach among others that will be mentioned within the protocol for determining the wedge factor. If you go off-axis, this is data from, from our group that shows that you can also model those as well analytically as well as virtual wedges that, that have a very steep off axis dependence but can be modeled fairly easily. Finally I wanna talk about off-axis ratios and there's a couple of methods to determine

points off-axis in photons beams, you can use off-axis dosimetry functions, there were a number of early papers that talked about doing that, that's not consistent within the protocols set up so we recommend the use of central axis dosimetry functions with sort of an all encompassing off-axis ratio which incorporates all the other effects including change in spectrum, etc. Those have to be determined either by measured data either through large field profile data, that's probably the simplest approach, most cases that's okay but if you get too far distances off-axis or deeper depths where there's a lot of scatter then that sort of breaks down. Primary off-axis ratios works quite well and will be recommended within the protocol along with situations of how to measure those. Analytic equations also can be used, this was a, a comparison that was done almost ten years ago that showed that for a number of different situations large field profiles were usually

within two percent only in extreme cases did they get out but in some cases they could be very far off, but primary off-axis ratios worked quite well. So with that I'm going to switch over to electron calculations. (Inaudible). Okay, thanks John, let's see, not too much time left, I'm going to try to buzz through electron calcs quickly, oops, wrong one. Okay, no. This one, okay. Back to nomenclature, some of the independent variables that we're gonna deal with with electron calculations are as follows; applicator field size, most accelerators would say have five applicators that you can use to collimate the beam, effective field size akin to phantom scatter factor so it's the field size on the patient surface after further collimation. G is a, a variable that's basically a gap factor, the distance between the treatment SSD and your normalization SSD which for our purposes here we'll call 100 centimeters. So if you have to go to an extended SSD,

you're going to get a gap. Effective SSD, the, the very bottom one, is an effective source to surface distance and I'll talk about how to, how we use that in just a second. A couple of dependent variables, there's an air gap correction factor which is used in a different method to calculate monitor units and extended SSD's, I'll talk about that in a second as well, and overall the electron output factor which most of you are familiar with in calculating electron monitor units and its certain all encompassing output factor. Okay, so the general equation that we use for this is the familiar equation that most of you are familiar with and that's the dose of prescription and I'm not gonna consider the case here where your prescribing to a percentage depth dose, I'm talking about dm here alone, divided by the output of the monitor units per, centigrade per monitor units and then also that output factor we talked about just a minute ago as

a function of the applicator and the let's call it, the effective field size here or the collimated field size. For extended SSD's, for example posterior neck electrons where you perhaps can't get your 100 SSD at the patient surface, there's two different methods that TG-71 is going to recommend akin to what TG-25 did. In this case you see the familiar factors here, but then there's this essentially inverse square factor down here which has effective SSD's plus depth and in this case effective SSD plus depth plus the gap that you had to use to, to treatment the patient and then a square term on the bottom there. The second technique from MD Anderson uses slightly differently in TG-71 the, the nominal SSD so 100 SSD whereas TG-25 that was a virtual SSD but for our purposes here we're gonna use the nominal SSD plus depth then the gap and then sort of a fudge factor here, which is the difference between a straight inverse square

attenuation and what the actual dosimetry shows. Okay, so as usual for square fields we're gonna measure the output factor during commissioning and if you have some extra time you could even measure those at different SSD's, but assuming you only do them at 100 SSD, that would be the standard data set that you get with a, with an accelerator. For rectangular fields we're gonna recommend that you use the, the square root method from Mills et. al., basically says that the output factor for a applicator field size in a rectangular field a length and a width is basically the square root of the product of the output factors of squares of those two sides. And many irregular fields can also be approximated by rectangular fields, here's a couple of examples, here posterior neck electrons, it's essentially a rectangle here so you can make that into a rectangle and just assume the dosimetry is the same. Some of these are a little more

complicated for example with the little wing out here, we suggest you treatment more as a rectangle that way assuming that a lot of the scatter from out here is not making to the prescription point anyway. Okay, so if the fields are very small however, you have to make some special considerations and what we consider a very small field is an effective field size or let's say a phantom field size that is smaller than the energy over two and a half. So for example, for a six centimeter field, if you're effective field size on the order of two or two and a half centimeters, then there's some special dosimetry that you have to consider because you're not getting equilibrium for example on that field. For bigger, for higher energy such as 20 MeV, now you're starting to talk about a 10 by 10 field but if you have an 8 by 8 field for example, and 20 MeV, you still have these problems with lateral scatter equilibrium. And so under these conditions, you can

determine your output factor by a couple of different ways, one is just the standard special dosimetry go out to you machine, compare the output of this particular setup, this particular clinical set up to a set up you that know the answer to, standard. And then also you can use the method of lateral build-up ratio by Khan that I, I'm gonna talk about perhaps just a touch here, but we're running low on time. The, the point is, that the smaller the field the, the more different the percent depth doses are so in these cases for a 6 MeV, 12 MeV and 20 MeV electron field, you can see what the percent depth doses look like for fields form small to large and this large is approaching lateral scatter, scatter equilibrium. So not only do you have different depths of the 90 percent for example, but

also you have different depths of D-max so under these circumstances, lots of things change. Not just the output, but also the depth of dm,

the depth of d90, the profile, so all those things need to be relayed to your physician so they know what they're treating under those conditions. Okay, just very quickly on this lateral build-up ratio, it's, it's the same thing that we do normally with an output except for the fact that you now have this thing called a lateral build-up ratio and a couple of factors that compare the fluence of an applicator to the fluence of a, of a reference applicator. Also the fluence of an insert, so you're cut out actually in this case to a reference insert, let's say, now I refer you to his paper published I think two or three years ago now to see what, what some of these actually come out to be. The lateral build-up, build-up ratio itself however, is the, as you can see down here, is the ratio of the central axis depth dose in a given circular field to the central axis depth dose in a broad field so in other words, with lateral scatter equilibrium for the same incident

fluence so basically you normalize at the surface and you can measure these data. You can take very, very small fields, you know, a radio field of one centimeter, two centimeters, three centimeters, essentially getting all the LBR's at commissioning time so you can determine your MU's down the road. So for extended SSD's, again, there's two different ways we can do these, they're required as you know in a lot of different cases for example posterior neck electrons, perineum, those kinds of things where you just can't get 100 SSD on the patient so you need to account for that. In using the, the air gap technique, you can determine this air gap factor by using just a straight inverse square correction with this virtual SSD and making up for the difference in the output you see as this air gap factor or again now we're not using virtual SSD anymore, we're using actual SSD, 100 SSD. Or you can use a straight inverse square correction

with no other term by determining what the effective SSD is and this is from a very old paper by Kahn but it shows you that if you measure at different SSD's what the output is versus what the output at 100 SSD and you plot the square root of the inverted ratio of those two measurements versus the gap itself, you get a line, and the best fit line, the slope of the best fit line, the inverse slope of the best fit line minus the reference depth gives you this thing called an effective SSD. And if you look at a table of these effective SSD's which perhaps you can't see, what happens is once you get to broad field, so anything over about 10 by 10, it doesn't, for, for energy, for field size, it doesn't really make a difference. The effective SSD is pretty much the same. As you get to lower energies and smaller field sizes, the effective SSD can change dramatically even over, you know, four by four to a six by six field to the point where you have extreme SSD effective

SSD's. Okay, let's move on from there. Talk briefly about quality assurance, monitor unit quality assurance. We're gonna strongly recommend that you have some independent verification of the monitor units that come out say, your treatment planning system for example if you have a treatment planning system that gives you MU's, you can use an independent treatment planning system to confirm those monitor units, you can have a stand alone program, either one that's commercially available or one that's

sort of a home built program or the old standard, you can look all the factors up in tables to make sure that your monitor units make some sense. We're gonna recommend that you follow Task Group 53's report on treatment planning systems such that you confirm that the input is the same, that the treatment planning system is getting the appropriate input that the algorithm makes sense for calculating monitor units and that it handled

heterogeneity, heterogeneities appropriately for monitor units for example you get, you get the appropriate depths if you're going through lung. Some of the vendors these days calculate monitor units in ways that we can't independently verify with the parameters they use so it's important we demand as users the appropriate data for example an effective depth or an effective field size such that we can do an independent calculation. And then it's also important for us to set a appropriate criteria; what is a, what is good enough agreement for our hand calculations, say to what the treatment planning system says for example a straight on spinal field should calc-, should agree probably much more accurately than for, for example a tangential photon on a breast patient. We're gonna also recommend that quality audits occur, the standard quality audit is the RPC sending you TLD's, again that's only, that's about as standard a condition as you can

get to, to irradiate a TLD with 300 centigray with no other associated parameters. You can also, however, get a thorough test set of different type of setups to make sure that your MU's are okay or you can hire an independent physicist to come in and look at your whole setup and make sure systematically that, that your MU's are correct. And finally, here's our conclusions. Task Group 71 of the Radiation Therapy Committee was formed to create a consistent nomenclature and formalism for MU calculations. For photons, we're gonna recommend a normalization depth of ten centimeters, although the formalism is valid for the maximum depth of maximum dose so if you choose not to go to ten centimeters, this formalism will still work. And for electron beams, we're gonna, we're going to allow for both effective SSD or air gap correction methods for extended SSD calculations. Thank you very much. Let's see, we have about five minutes for

questions. Q: When this task, task group report will be out? A: We are, we've been extended on our deadline which we have a draft in, we need to have comments back from the Radiation Therapy Committee reviewers, and let's hope that we'll have a more or less final report by the end of the year. Q: The a, your showing that the two jaws, the upper and lower jaws, give different collimator or output factors and that if you normalize them to how much the flattening filter exceeds the square ones and each jaw all line up very nicely. That seems to me to say that I can probably just if I do what one implication of yours with the table is, I can get a textbook worth of the 1 by, 1 by 20, 2 by 20, 3 by 20, or 4 by 20, you know, all that a lot of output factors or I can just use the, the square method where I just multiply sort of a square lower jaw times the square upper jaw times square root, maybe even play with A___ jaws the same way. Would that, does

that make sense what I'm suggesting? A: Yeah, well, the, the one, one thing to emphasize first is that there's usually not a whole lot of difference between the two jaw sizes so, and the jaws are pretty close to each other so the rectangular field shape is pretty close to the square and the corresponding equivalent square that you would've already measured. So

the effect is usually not that large if you just use the equivalent square that you would use by looking at it at 100 centimeters, you're gonna be close. If you do wanna go closer, it's really a geometry issue within the point's eye view model. Depends on where the jaws are, where the flattening filter's located relative to the isocenter and within the protocol within the appendix there are equations for determining exactly what the, how to convert from a particular jaw setting to the field size at the flattening filter in which case you can then look that up on your equivalent square table, which is

much easier than measuring that huge amount of data. Q: A couple of comments on your formalism, it, to do with the normalization point, the normalization point is just the point that you tie your system together at so as long as you adopt a consistent point, it's not going to affect your accuracy and in fact, the example you gave where you showed the difference in the TPR tables for the different field size settings, you're still going to be as inaccurate if you calculate at D-max for a large collimator setting and a small exposed area so it actually gives you an incorrect answer no matter how you tie your dosimetry system together in that situation. The second point is that, when we go to a normalization condition and we tie all everything together, since the normalization point is an arbitrary factor, I've found that actually tying it in air and combining my PS and TPR together, actually makes for a much simpler calculation in that you're just not looking

up as many factors 'cuz you very rarely will look up a TPR without looking up the exact same PS factor at the same time. A: That the, that's, that's correct and the, the, the latter comment is correct. You could combine those if you wanted to, they both have the same field size dependence for isocentric calculations not for percent depth dose because technically the field size for the phantom scatter for percent depth dose is the field size at the reference depth and regards to your first comment, that particular scenario that was shown here, that showed the problems with the D-max formalism, is really only for isolated cases. We're talking about high energy photon beams, we're talking about cases where the collimator scatter, the collimator is opened significantly more than the blocks and the reason that you get that problem is it's all, you're just talking about normalization points, but when we normalize to a depth of D-max, you're

relating everything out back to say, for example, one centigray per monitor unit for a 10 by 10 field size. Everything is set for 10 by 10 and now, if I wanna calculate the dose to depth of five or depth of ten, if you have a reference depth of ten, or I'm sorry, depth of five or some other deeper depth, we typically don't calculate dose to D-max. We're referencing a depth of, arbitrary depth of D which is not a D-max to a depth of ten. Where we don't have that problem when we open the collimator jaws that we have added electron contamination which influences our TPR, but if you're referencing back to D-max, you do. So it's a little bit of an extreme situation, it's not going to happen very often, but you can have some problems with D-max that way. Q: Yeah, but that, that's not, I mean, that's the intermediate step in the calculation. As long as your tables are consistent, you're not going to have a problem there. A: Could, could I, could I get in here for a

second, I mean, I think the point is, is that do you want the inaccuracy to exist at shallow depth, you're, you're right, the inaccuracy is the same but when your normalization is at D-max it means D-max is your most accurate point and your error propagates to the clinically relevant depths whereas if you choose depth of ten, your most accurate now at ten and your error's localized to D-max and I think most of us would agree that we, we take an inaccuracy at D-max, if we can get accurate doses over the clinically relevant range from say five to 20 as opposed to having, you know, very accurate monitor units at D-max and having our error at the clinically relevant depths of five to 20. Q: Yeah, I'm not, I'm not saying we should normalize or we should measure at, at D-max, I mean I completely agree that we should be measuring things at ten centimeters depth but my point is that in, if we are norm-, if we do the calculation back to D-max, if we use a consistent

dosimetry system, it doesn't matter where we have our normalization condition. So. A: It does, actually. Q: No, no it won't. It'll all, it should all work out because if you think about it, in your other table when you showed that TPR and you normalized everything to D-max, those curves would then come into agreement except up at D-max and when you go back and if you're doing your calculation near D-max, this is the situation. If you're doing a calculation near D-max, you, you're, you'll see the same discrepancies coming out when you calculation in your formalism. So you're not really saving yourself anything and that's the point. A: Well, I think we can probably continue this discussion after the, the talk. Are there any other questions? Q: Yes, I have a couple of questions here. The first question is about photons, you mentioned about the primary off-axis ratio, would like if you can give me some details about that 6.7 percent that you were

mentioning where you recommend to use that instead of the normal OAR. That's one question. The second one is about electrons, if you, if the protocol address the obliquity factors for electrons, this no, I didn't see any kind of recommendations so I want to ask about that topic also. A: Okay, first of all, to answer the lighter question, sorry, with the, the, the data which was shown, those values that had a max-, that was the maximum error and that was associated with a 15 centimeter off-axis field, at a, at a depth of 15 centimeters so it's an extremely high off-axis and the reason that you have that problem is that when you use measured profile data, when you get down to that depth, you have a significant component of scatter that's contributing to all of the dose there and when you go very far off-axis, that has a large reduction in scatter from what you have on the central axis so the amount of scatter contribution which is implicit in the off-axis

profiles, is significant down at those depths. So when you're calculating dose and you're trying to determine what the primary dose is to that other point, if you use an off-axis profile for measured profile data, then you, you're not accounting for the fact that you've lost a lot of scatter in the profile and you're actually underestimating the off-axis ratio by quite a bit. Q: Thank you. Electrons? A: Well, for the electrons we don't address obliquity in the report. Q: I have one question concerning effective SSD for the electron, you showed a very nice table from Medical Physics 1995 which shows that at least for the Varian____, the effective SSD is function more of the cutout area than the cone. (Right.) Do you happen to know whether this is true, in general, for other type of ____ or

is this just a property of a Varian _____ for their, for their electron beams? A: I, can I comment on that? Absolutely. That's typically a function of whether or not the

manufacturer of the applicator's relying on the applicator to actually provide additional scattering into the field to compensate for flatness. For example, the Varian applicator is very well designed to give you very flat profiles which means a significant amount of your source of electrons is actually coming from the applicator and therefore if you consider like an electron fluence, it's very much effective 'cuz you sort of have, like, this source up here and a source that's really part of the applicator itself. Whereas on the old Siemens units, not the, not the more monitor ones, but the old Siemens set of applicators, that really wasn't designed in and you would find very little of this effect as a, as a function of the field size because the applicator itself wasn't providing a significant portion of the electrons that were contributing dose. So it's clearly a, a result of the design of the applicator and something you have to evaluate on a machine by machine basis.

John, I think I'd like to just take this opportunity to thank TG-71, we've been fighting with this document for in excess of 20 years, and it looks like maybe we're there finally and I think perhaps we may get to a point where physicists and dosimetrists moving from treatment planning computer to treatment planning computer and institution to institution may in fact be able to avoid the confusion that always occurs because of the wide diversity in dosimetry programs so, thanksto TG-71. I really think we all need to give you a large round of applause.