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Direct megavoltage photon calibration service in Australia

D. J. Butler,

Australian Radiation Protection and Nuclear Safety Agency, 619 Lower Plenty Road, Yallambie, VIC 3085, Australia

G. Ramanathan,

Australian Radiation Protection and Nuclear Safety Agency, 619 Lower Plenty Road, Yallambie, VIC 3085, Australia

C. Oliver,

Australian Radiation Protection and Nuclear Safety Agency, 619 Lower Plenty Road, Yallambie, VIC 3085, Australia

A. Cole,

Australian Radiation Protection and Nuclear Safety Agency, 619 Lower Plenty Road, Yallambie, VIC 3085, Australia

J. Lye,

Australian Radiation Protection and Nuclear Safety Agency, 619 Lower Plenty Road, Yallambie, VIC 3085, Australia

Australian Clinical Dosimetry Service, 619 Lower Plenty Road, Yallambie, VIC 3085, Australia

P. D. Harty,

Australian Radiation Protection and Nuclear Safety Agency, 619 Lower Plenty Road, Yallambie, VIC 3085, Australia

T. Wright,

Australian Radiation Protection and Nuclear Safety Agency, 619 Lower Plenty Road, Yallambie, VIC 3085, Australia

D. V. Webb, and

Australian Radiation Protection and Nuclear Safety Agency, 619 Lower Plenty Road, Yallambie, VIC 3085, Australia

D. S. Followill

Radiological Physics Centre (now the Imaging and Radiation Oncology Core Quality Assurance Centre), M. D. Anderson Cancer Center, 1515 Holcombe Blvd, Houston, TX 77030, USA

D. J. Butler: Duncan.Butler@arpana.gov.au

Abstract

The Australian Radiation Protection and Nuclear Safety Agency (ARPANSA) maintains the Australian primary standard of absorbed dose. Until recently, the standard was used to calibrate ionisation chambers only in ^{60}Co gamma rays. These chambers are then used by radiotherapy clinics to determine linac output, using a correction factor (k_Q) to take into account the different spectra of ^{60}Co and the linac. Over the period 2010–2013, ARPANSA adapted the primary standard to work in megavoltage linac beams, and has developed a calibration service at three photon beams (6, 10 and 18 MV) from an Elekta Synergy linac. We describe the details of the new calibration service, the method validation and the use of the new calibration factors with the International Atomic Energy Agency's TRS-398 dosimetry Code of Practice. The expected changes in absorbed dose measurements in the clinic when shifting from ^{60}Co to the direct calibration are determined. For a Farmer chamber (model 2571), the measured chamber calibration coefficient is expected to be reduced by 0.4, 1.0 and 1.1 % respectively for these three beams when compared to the factor derived from ^{60}Co . These results are in overall agreement with international absorbed dose standards and calculations by Muir and Rogers in 2010 of k_Q factors using Monte Carlo techniques. The reasons for and against moving to the new service are discussed in the light of the requirements of clinical dosimetry.

Keywords

Radiotherapy; Dosimetry; Absorbed dose; Protocol; TRS-398; Megavoltage calibration

Introduction

Dosimetry for external beam radiotherapy in Australia is traceable to the primary standard of absorbed dose, a graphite calorimeter of the Domen type [1,2], operated in a beam of ^{60}Co radiation [3]. This standard is maintained at the Melbourne laboratories of the Australian Radiation Protection and Nuclear Safety Agency (ARPANSA) under an authorisation from the National Measurement Institute (Australia). Radiotherapy facilities use the calibration service in combination with a calculated correction factor, k_Q , tabled in the IAEA TRS-398 Code of Practice [4], to measure linac output. In Australia and New Zealand, secondary standard laboratories can also provide a ^{60}Co calibration which can be used with TRS-398. In this way, all patient treatments delivered using external beams in this region are traceable to the calorimeter operated in ^{60}Co .

ARPANSA has recently developed a calibration service to allow most therapy ionisation chamber types to be calibrated directly at three megavoltage photon energies: 6, 10 and 18 MV. The new service is more accurate than using a calculated k_Q since the chambers are calibrated in beams very similar to those in radiotherapy clinics. The accuracy of a calculated k_Q factor depends on the precision of the chamber geometry (including any deviations from the specified geometry) and the accuracy of the calculation. To use a *measured* k_Q only requires that the chamber response be reasonably insensitive to photon energy, so that users can interpolate with the beam quality index. ARPANSA provide a quadratic fit to the calibration coefficients as a function of the beam quality $\text{TPR}_{20,10}$ in the calibration report for purposes of interpolation. The new service is based on the same primary standard graphite calorimeter adapted to work with linac beams. Adaptations were

required to account for variations in dose rate during irradiation and between irradiations, and for the higher dose rate compared to ^{60}Co . Additional graphite buildup was also employed [2, 3].

The purpose of this article is to explain the fundamental differences between these two methods of traceability for absorbed dose in linac photon beams (via ^{60}Co or via “direct” calibration). We present the validation of the new method, and compare the uncertainties in both cases. We present the change in absorbed dose expected in the clinic when shifting between the methods. We include a detailed discussion of the advantages and disadvantages of these methods from the radiotherapy users’ point of view. In particular, we discuss the two services in the light of the expected shift in dose, the origin of dose prescriptions, the improvement in dose accuracy which comes from the new service, and the expected shift in patient doses in Australasia compared to the rest of the world.

The new service is available outside Australia and could be used by New Zealand clinics. The standards laboratory of New Zealand, the National Centre for Radiation Science, has taken part in tests of the service. The pros and cons of this article are relevant, with the additional drawback of having to ship dosimeters overseas.

ARPANSA plans to extend the new method to mega-voltage electron beams. However at present electron dosimetry remains traceable to the primary standard realised on ^{60}Co —either through cross-calibrations performed in the clinic, or by using the new cross-calibration service at ARPANSA. In both cases a parallel plate chamber is calibrated on a high energy electron beam against a thimble chamber which has been calibrated in ^{60}Co .

Review of calibration services outside Australia

The primary standards laboratory of the UK, the National Physical Laboratory (NPL), introduced a calibration service for megavoltage photon beams in 1989 and for electron beams in 2001. These calibrations are used with the UK Codes of Practice [5, 6] as the basis for reference dosimetry in UK clinics.

In North America, the primary standards laboratories of the US and Canada, the National Institute of Standards and Technology, NIST, and National Research Council Canada, NRCC, respectively, maintain linear accelerators and are able to calibrate ionisation chambers directly in terms of absorbed dose at linac energies. They do not routinely provide calibrations to all radiotherapy facilities, however, as the large number of linacs in North America (more than 5,000) requires the existence of a network of secondary standards laboratories. These laboratories (Accredited Dosimetry Calibration Laboratories, ADCLs), do not have easy access to linacs, and presumably a business case weighing the relatively small improvements in accuracy brought about by using linacs is difficult to make when the reliability and lower cost of ^{60}Co is taken into account.

The NRCC tests representative chambers to ensure that those calibrated via ^{60}Co do not result in clinical dose errors. For example, in 2010 NRCC published results of measurements of a wide range of chambers indicating which chambers were suitable for reference dosimetry [7]. Megavoltage external beams in North America are calibrated according to the

AAPM protocol TG-51 [8] which only allows ^{60}Co as the reference quality. A review in 2014 continued this recommendation [9].

Across Europe, most ionisation chamber calibrations are performed in ^{60}Co . In France and Germany, the primary standards laboratories Laboratoire National Henri Becquerel (LNHB) and Physikalisch-Technische Bundesanstalt (PTB), respectively, maintain primary standards on both ^{60}Co and medical linacs. The majority of clinical dose measurements are traceable to chambers calibrated in ^{60}Co [10]. In Belgian and Dutch hospitals [11], chambers are calibrated in ^{60}Co and used with a local protocol NCS-18 [12] which includes k_Q factors derived from measurements made in a representative set of chambers at several hospitals using portable water calorimeters.

In Japan, Korea and Taiwan, dosimetry is currently based on ^{60}Co , but the standards laboratories have installed linacs and are in the process of setting up calibration services. Whether the clinics use these services directly, or whether the services are used to check ^{60}Co based dosimetry, remains to be seen.

In summary, nearly all external beam radiotherapy around the world is traceable to primary standards of absorbed dose realised on ^{60}Co and occasionally MV linacs, using a variety of protocols.

Cobalt-60 versus linac calibration service

In Australia, nearly all external beam radiotherapy is delivered using linacs with accelerating potentials in the range 4–18 MV [13]. Beam output dose measurements are performed with either a Farmer chamber (model 2571) or a chamber of similar design (“Farmer-type”). The PTW 30013 and IBA FC65-G are the most common chambers in addition to the 2571. As mentioned, these chambers are calibrated for absorbed dose to water in ^{60}Co but ARPANSA can now calibrate them directly in linac beams. Here we consider the differences in the two calibration chains and in the results.

Firstly, ^{60}Co gamma rays are continuous, have relatively low dose rate in radiotherapy, and have a lower average energy than medical linac beams. These differences are summarised in Table 1.

Using a calculated factor to take into account the different response in a linac beam is therefore an extrapolation which involves assumptions about the dosimeter. The ionisation chamber geometry is assumed to be the same as that used for the calculation of k_Q . The response of the electrometer to a pulsed current is assumed to be the same as for a constant current. These assumptions have been shown to hold nearly all of the time for modern radiotherapy dosimeters [7]. Nevertheless, there is always the risk that a chamber or electrometer does not conform to these assumptions, either because it is damaged, has been manufactured differently, or the design is flawed.

In a direct calibration, the response of the dosimeter is determined by interpolation of k_Q for different $\text{TPR}_{20,10}$ values. The beam qualities, dose rates and pulse behaviour are

qualitatively similar for all medical linacs. No knowledge of the chamber construction or behaviour is required, and almost any chamber type can be calibrated.

The second, related point to consider is the uncertainty in clinical dose measurements derived from each method. Here we intend 'clinical doses' to refer to all doses delivered by the clinic, i.e. patient doses and doses delivered under reference conditions. Using ^{60}Co with factors from TRS-398, for example, the uncertainty in the calibration coefficient of a Farmer chamber at megavoltage photons is around 1.1 % (standard uncertainty $k = 1$). Using the direct calibration method this can be reduced to around 0.6 %. Clinical doses determined via the direct calibration route should therefore be more accurate (although we note that the use of more accurate Monte Carlo calculated k_Q factors such as those recently included in TG-51 [9] would reduce this improvement).

While the absolute accuracy of dose measurements will increase with the direct calibration service, it is less clear that the overall consistency in clinical doses will also increase. That is, random contributions to calibration factors on a linac are likely to be worse than for ^{60}Co where the beam is more reproducible. The agreement between standards laboratories of different countries is slightly worse for linacs than it is for ^{60}Co , as can be seen in the results from the Key Comparison Database [14] (accessed in February 2014) (Table 2). This database records the results of comparisons between standards laboratories around the world and the central primary standards laboratory, the Bureau International des Poids et Mesures (BIPM). The standard deviation of the ratios of absorbed dose to water for the BIPM.RI(I)-K6 comparison (linacs) for four laboratories and three photon beams is 0.0038 (0.38 %). The same ratio for ^{60}Co absorbed dose to water in BIPM.RI(I)-K4 (for ^{60}Co) is 0.0020 (0.20 %) for thirteen laboratories (Table 2).

Other differences between ^{60}Co and linac based calibrations are practical. A ^{60}Co unit costs much less than a linac. The linac has higher maintenance costs and presumably greater down-time, but also a potentially longer life, as ^{60}Co can only be maintained for around 10 years before the source decays to below a usable level. The linac also requires more detailed quality assurance measurements to make sure the beams are symmetric and the energy is constant, while ^{60}Co requires physical security measures and presents difficulties in transport and disposal at its end of life. We estimate the cost of the linac to around ten times that of ^{60}Co . Nevertheless, the linac brings many capabilities beyond calibrating ionisation chambers (research, development of radiotherapy dosimetry auditing procedures, liaison with the medical physics community) which should be considered in any cost-benefit analysis.

In ^{60}Co , the calibration procedure uses the source output measured by the primary standard and corrected for source decay. An ARPANSA ionisation chamber is used to check this rate during the calibration. The direct calibration service uses a secondary standard chamber, against which the user chamber is cross-calibrated at three energies. The total amount of time for each calibration (^{60}Co or 3 linac beams) is of the same order.

Validation of the new service

Before introducing the new calibration service, ARPANSA undertook three international comparisons and a literature survey of measured k_Q values. ARPANSA also conducted a field trial to test the service, seek feedback, and to allow enough time to establish baselines for quality control.

Comparisons with other measurements of absorbed dose in linac beams

In September 2012, a team from the BIPM visited ARPANSA with a primary standard graphite calorimeter to perform absorbed dose measurements on the ARPANSA linac. These measurements were compared with ARPANSA's measurements and the resulting comparison was published recently [15]. ARPANSA also took part in two indirect bilateral comparisons where ionisation chambers were measured at ARPANSA and in another laboratory. These comparisons are less direct because the chambers are calibrated on different linacs and some interpolation of the results is required. The results showing the ratio of the ARPANSA dose to water calibration coefficient to the other laboratory is shown in Table 3 and Fig. 1, for all three comparisons. Most of the ratios fall within one standard uncertainty of unity, indicating reasonable agreement. The agreement is worst for the highest energy beam, but still within two standard uncertainties.

Comparison with measured and calculated 2571 k_Q values

As a second form of method validation, we compare the k_Q factors for the 2571 chamber type obtained by ARPANSA with published values. This factor is known to vary by only a small amount from chamber to chamber [7]. Measured k_Q 's are plotted in Fig. 2 for the Netherlands [12], Canada [7], UK [12], France [12] and ARPANSA. Also shown are the calculated values from TRS-398 [4] and the more recent Monte Carlo calculations of Muir and Rogers [17]. The results are consistent with the comparisons results given in the previous section: ARPANSA is lower than the average but agrees within the spread of international values.

Uncertainties

The uncertainty in the user chamber calibration coefficient is given in Table 4 (based on a calibration in ^{60}Co and use of TRS-398) and Table 5 (based on calibration in the three ARPANSA reference linac beams and interpolation with $\text{TPR}_{20,10}$). The uncertainty in the calibration coefficient is significantly smaller for the direct calibration method.

Use of direct linac calibration service with TRS-398

The TRS-398 dosimetry protocol recommends the use of direct calibrations where available, over ^{60}Co , for both megavoltage photons and electrons. Our uncertainty calculation indicates that the resulting clinical doses will have lower uncertainties than those based on ^{60}Co .

Section 6.5.2 of TRS-398 concerns the use of a measured k_Q . It recommends that the user interpolate to the user energies. A method of interpolation is not specified, but ARPANSA provides a simple quadratic fit. It also recommends the use of reference quality Q_0 and

normalising other calibration coefficients to this value to produce a set of k_{Q,Q_0} correction factors. ARPANSA has adopted this recommendation and uses ^{60}Co for Q_0 .

Shift in absorbed dose when adopting the new service

Of particular interest to radiotherapy clinics is the shift in patient dosimetry that would accompany adoption of the new service. While clinics will welcome the reduction in uncertainty that comes with the new service, they will also be mindful of time lost checking discontinuities in quality control data and the need to re-establish some baseline measurements.

Figure 3 shows the ratio of the factor k_Q measured by ARPANSA, to that tabulated in TRS-398, for the three most common chamber types in use in Australia and for three representative beam qualities (nominally 6, 10 and 18 MV). The measured k_Q 's are the average of two different chambers of each type.

The differences shown in Fig. 3 will translate directly into differences in clinical linac output measurements. For example, a clinic using a 2571 chamber should expect a dosimetry shift of nearly 0.5 and 1.2 % at 6 and 18 MV, respectively. The shift at all energies will lower $N_{D,w}$ values compared to those based on ^{60}Co . The consequence of this shift is that linacs will be calibrated to deliver *more* radiation when based on dosimetry from the new (lower) $N_{D,w}$ coefficients. The magnitude of the shift is within the combined standard uncertainties of the two calibration methods, however the change is not insignificant from a user perspective.

Discussion

Traceability and consistency in radiotherapy dosimetry are important for several reasons. We discuss these issues with regard to the different uncertainty and shift in dosimetry which accompanies the new calibration service.

Dose accuracy requirements

Differences in clinical doses which arise due to differences in traceability are usually not reported. A recent article by Andreo is an exception [19]. Nevertheless, improvements in the consistency and accuracy of ionisation chamber calibrations are considered to be important in the clinic. For example, dose errors of more than 5 % have been shown to have measurable effects on patient outcome for some treatments [20]. This implies that ionisation chambers need to be calibrated to better than 2.5 % standard uncertainty if all other contributions to the clinical uncertainty are zero, and much more accurately than this if clinical delivery uncertainties are taken into account. Boyer and Schultheiss [21] conclude that 1 % in dose correlates with about a 2 % change in early-stage tumour control. A more recent analysis by Thwaites et al. [22] concludes that 3 % standard uncertainty in the dose delivered to the patient can be taken as the “currently recommended general accuracy requirement”. Further discussion can be found in references [20–23]. These arguments support the need for more accurate absorbed dose standards.

The origin of clinical dose prescriptions

Radiotherapy prescriptions are in most cases derived from clinical trials. A wide variety of dose fractionation schemes are employed for the same clinical scenario [24, 25]. Sometimes the different prescriptions arise from different intent (the level of complications considered acceptable versus the degree of tumour control, for example), other times because they are based on different trials, or different historical practices and experience.

In all of these trials, the traceability of the dosimetry in the trial is difficult to extract, and is considered to be below the threshold that would influence the trial results. Dosimetry based on air kerma standards in ^{60}Co is known to result in linac outputs which differ by up to a percent when compared to dosimetry from ^{60}Co absorbed dose to water standards [26, 27]. Differences of this order between countries are also possible because the primary standards can differ by this amount, even absorbed dose standards. It is therefore not possible to argue for or against the use of ^{60}Co from a prescription point of view.

Absolute dose accuracy and treatment repeatability

The ability to measure absorbed dose to water in absolute terms is required for radiotherapy only because it is the quantity chosen for prescriptions. Prescriptions need to be made in a quantity that (a) can be realised with a small uncertainty, (b) can be measured consistently and internationally, and (c) has a response which is proportional to tissue and tumour response. Absorbed dose to water is used mainly because it fulfils these criteria, but its relationship to tissue response needs to be established by clinical trials. It is possible to conceive of more appropriate quantities (such as the number of double-strand breaks per cell) which might be more proportional to tissue response. It is also possible to conceive of quantities that could be realised more consistently around the world (such as the ionisation in an air filled chamber). We make this comment to note that improving the accuracy of absorbed dose standards is not necessarily what is required by the radiotherapy profession. For example, radiotherapy doses might be more consistent if based on ^{60}Co and agreed values of k_Q were adopted and prescribed for use, even if the absorbed doses were incorrect. We believe, however, that the best way to improve consistency in radiotherapy is to improve the measurement of absorbed dose. By basing radiotherapy on a physical quantity, dose prescriptions will always have a physical meaning. This will not only help prevent misunderstandings with existing treatments, but will ensure that new treatment modalities have a framework for dosimetry.

Clinical doses in Australasia versus North America

In addition, we can look at linac output in Australia and New Zealand compared to the rest of the world. The Radiological Physics Centre (RPC) [28] in Houston regularly conducts remote audits of linac output around the world. Their results for photon beams audited in Australasia between 1999 and 2013 are shown in Fig. 4. These audits used TLDs up until 2010 and OSLDs afterwards. The same data for North America have a mean of 0.998 and standard deviation of 0.015. The mean of the Australasian data is 0.989 with standard deviation 0.015 from 448 measurements. The standard deviation of the mean is 0.07 % (less for North America) which indicates that the results are statistically significant: audited Australasian clinical doses are slightly lower than in North America. Differences in

dosimetry protocols and primary standards in Australasia and North America account for some of the difference. Other differences, such as reference temperature of 22 °C in North America versus 20 °C in Australia, and potential differences in reporting dose to tissue verses dose to water, are not expected to have an effect. In 2013, a comparison conducted with a single ionisation chamber between the RPC and ARPANSA showed only 0.16% difference at ^{60}Co . Nevertheless, based on these audit data, we can expect that Australian linacs will be closer to those of North America if the output is increased by about 1 %.

Monte Carlo k_Q factors

A final comment on the adoption of the direct calibration service concerns the accuracy of the calculated k_Q factors in TRS-398. These are known to differ from recent measurements and calculations by a small amount (within their stated uncertainty) [19], mainly because the chamber stem was not included in the calculation of k_Q . When Monte Carlo calculations are used to derive k_Q factors and the stem is included, the results are in better agreement with ARPANSA's and other laboratories' measured values. Figure 5 shows the Monte Carlo calculated k_Q factors calculated by Muir and Rogers [17] for the most common chambers used in Australia. These values are also included in Fig. 2 for the 2571 chamber. The results indicate that at least some of the difference between TRS-398 used with ^{60}Co and the ARPANSA direct calibration is due to the TRS-398 k_Q factors.

Flattening Filter Free (FFF) beams

The use of the TRS-398 Code of Practice with FFF beams has been shown to introduce a small additional error of around 0.5 % [29], because the relationship between $\text{TPR}_{20,10}$ and stopping power ratios is different for FFF beams compared to flattened beams. This difference results in a different chamber response for the same $\text{TPR}_{20,10}$, depending on whether the beam is a normal flattened beam or an FFF beam. A chamber calibrated directly on flattened beams will therefore entail the same error on an FFF beam as using a chamber calibrated at ^{60}Co , when the beam quality index is $\text{TPR}_{20,10}$. Possible solutions include reporting the calibration results in terms of $\%dd(10)_x$, which has a better correlation with stopping powers, so that the measured k_Q 's could be used with TG-51 [8] for FFF beams, or ARPANSA could implement FFF beams on their linac and provide direct calibrations for these beams.

Conclusion

The ARPANSA calibration service for megavoltage linac photons has been described. The results of three comparisons and an indirect comparison of k_Q factors for the 2571 chamber indicate that ARPANSA dosimetry is within the estimated uncertainty of other international standards of absorbed dose. The shift in clinical dosimetry when using the service instead of ^{60}Co is expected to be in the range 0.5–1.3 %, with directly measured calibration coefficients being lower and clinical doses being higher, than for ^{60}Co , when using the TRS-398 Code of Practice over the range $\text{TPR}_{20,10} = 0.62\text{--}0.80$. The advantages of the new service have been discussed in the light of the clinical impact of changing the service, with the conclusion that the move is justified based on the improvements in dose accuracy.

Radiotherapy facilities can use this information to decide whether to shift to the new service when the additional cost is taken into account.

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Fig. 1.

Results of international comparisons of absorbed dose to water (Table 3) between ARPANSA and three other laboratories. The results for NRC and NMTJ have been interpolated to the ARPANSA beam qualities shown on this graph. A small horizontal offset was used to separate the data

Fig. 2.

Correction factors k_Q for the 2571 chamber type measured by different laboratories on medical linacs (The Netherlands [12], Canada [7], UK [12], France [12] and Australia). Values from TRS-398 [4] and Monte Carlo calculations of Muir and Rogers [17] are also shown. Standard uncertainties are shown for ARPANSA and TRS-398, the other points have uncertainties in the range 0.5–1.0 %

Fig. 3.
Ratio of the k_Q 's measured by ARPANSA to the values tabulated in TRS-398, for three common chamber types and three beam qualities

Fig. 4.

Ratio of the RPC-measured dose and clinic stated dose for remote output measurements, for all radiotherapy clinics audited by the RPC in Australasia. All photon beam audit results are shown. Repeat measurements and beam energy are not differentiated. Data courtesy of the RPC, Houston, USA

Fig. 5. Ratio of k_Q 's calculated using Monte Carlo techniques by Muir and Rogers [17] to the values tabulated in TRS-398, for five common chamber types, over all the $TPR_{20,10}$ values in TRS-398. Nominal accelerating potential ranges are shown as vertical stripes

Table 1Differences between ^{60}Co gamma radiation and MV linac X-rays (nominal values only)

Quantity	^{60}Co	MV linac X-rays
Spectrum	Two gammas plus scatter	Bremsstrahlung plus scatter
Nominal energy	1.3 MeV	1–25 MeV
Nominal D_w rate	1–10 mGy/s	60 mGy/s
Peak D_w rate	10 mGy/s	60,000 mGy/s
Type	Continuous	Pulsed
Output over a few minutes	Constant	Ramp up then usually constant to within about 3 %

Table 2

Results for key comparisons of absorbed dose to water in linac beams (three energy ranges and four laboratories) and in ^{60}Co (13 laboratories)

Lab, i	Country	TPR _{20,10}	Ratio of D_w (Lab, i/BIPM) x_i	Standard uncertainty in ratio u_i	Date of measurement
NRC	Canada	0.681	0.9973	0.0055	2009
PTB	Germany	0.683	1.0013	0.0052	2010
NIST	US	0.674	1.0035	0.0057	2010
LNE-LNHB	France	0.675	0.9952	0.0044	2012
NRC	Canada	0.731	1.0008	0.0055	2009
PTB	Germany	0.733	1.0034	0.0057	2010
LNE-LNHB	France	0.749	0.9948	0.0047	2012
NRC	Canada	0.800	0.9942	0.0055	2009
PTB	Germany	0.798	1.0018	0.0064	2010
NIST	US	0.783	0.9958	0.0059	2010
Standard deviation of linac comparison ratios $x_i = 0.37\%$					
NIST	US	Co-60	0.9984	0.0051	1997
METAS	Switzerland	Co-60	0.9999	0.0054	2000
MEKH	Hungary	Co-60	0.9983	0.0048	2001
PTB	Germany	Co-60	0.9961	0.0037	2005
VSL	Netherlands	Co-60	0.9926	0.0049	2005
ENEAC	Italy	Co-60	0.9999	0.0044	2007
NPL	UK	Co-60	0.9980	0.0064	2007
BEV	Austria	Co-60	0.9996	0.0044	2009
VNIIFTRI	Russia	Co-60	0.9976	0.0043	2009
NRC	Canada	Co-60	0.9980	0.0052	2009
NMIJ	Japan	Co-60	0.9960	0.0046	2009
ARPANSA	Australia	Co-60	0.9973	0.0053	2010
LNE-LNHB	France	Co-60	0.9971	0.0039	2013
Standard deviation of ^{60}Co comparison ratios $x_i = 0.20\%$					

Compiled from the online key comparison database [14], February 2014

Results of three international comparisons of absorbed dose to water (1) the Key Comparison BIPM.RI(1)-K6 ARPANSA and BIPM [15], (2) ARPANSA and NRCC (Canada) and (3) ARPANSA and National Measurement Institute (Japan) NMTJ [16]. R is the ratio of the absorbed dose to water realised by ARPANSA to the value realised by the other laboratory

Table 3

Nominal accelerating voltage (MV)	Measured $TPR_{20,10}$	Ratio R ARP/BIPM	$u_c(R)/R$	Ratio R ARP/NRCC	$u_c(R)/R$	Ratio R ARP/NMIJ	$u_c(R)/R$
6	0.673	0.9965	0.0055	0.9954	0.006	1.0000	0.007
10	0.734	0.9924	0.0060	0.9928	0.006	0.9972	0.007
18	0.777	0.9932	0.0059	0.9920	0.006	0.9930	0.007

Table 4

Uncertainty in the calibration coefficient of an ionisation chamber at the user quality Q when based on a ^{60}Co calibration

Source of uncertainty	u (%)
ARPANSA calibration coefficient $N_{D,w,\text{Co-60}}$	0.4
TRS-398 k_Q	1.0
Combined relative standard uncertainty ($k = 1$)	1.1

Table 5

Uncertainty in the calibration coefficient of an ionisation chamber at the user quality Q when based on calibration in the ARPANSA reference qualities Q_0

Source of uncertainty	u (%)		
	6 MV	10 MV	18 MV
ARPANSA calibration coefficient N_{D,w,Q_0}	0.44	0.49	0.49
Interpolation to user qualities	0.1	0.1	0.1
Spectral differences between calibration and clinical beam ^a	0.40	0.40	0.40
Combined relative standard uncertainty (k= 1)	0.62	0.65	0.65

^aBased on modelled differences in ionisation chamber response for beams of the same TPR_{20,10} but different spectra [18]