

A gauge for measuring the dose rate and activity of ophthalmic applicators

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Abstract A gauge was developed for determining the dose rate distribution and surface activity of ophthalmic brachytherapy applicators, particularly for ^{106}Ru applicators. A plastic $\phi 2 \times 2$ mm scintillator is used as the radiation detector, featuring a high pulse count rate, which results in a low 0.5% random error, due to good counting statistics. Automatic gain control of the photomultiplier tube (PMT) is achieved using a LED as the reference light source. The PMT operates in pulse mode. Long term gain variation due to fatigue of the PMT or ambient temperature variation is thus compensated for. The count rate error due to inaccurate setting of the high voltage supply of the PMT is 0.4%, and the instability error over 7 hours of continuous operation does not exceed 1–2%, peak-to-peak.

Key words dose rate • measurements • ^{106}Ru ophthalmic applicators

Introduction

In radiation oncology ^{106}Ru or ^{125}I brachytherapy applicators are commonly applied for treating eye melanoma tumours [1, 3]. The dosimetry of beta-ray sources for therapeutic applications is particularly difficult. Due to the high absorption of the beta radiation emitted by these sources, millimetre ranges and high dose rate gradients are involved. Another difficulty lies in the non-uniform distribution of source activity on the applicator itself, causing a highly irregular dose distribution. Brachytherapy applicators are usually of concave shape with a curvature radius of 10 to 15 mm and a diameter typically between 11 mm and 25 mm, to match the shape of the eye ball. In some applicators an opening is made for the visual nerve. Depending on the tumour localization and the size of the eye globe, an appropriate applicator is attached to the treated eye for a time period required to treat the volume of interest with a recommended dose. Modern diagnostic techniques permit a precise determination of all volumes of interest in the eye, i.e. that of the tumor and of the critical structures. It is, therefore, of importance to optimize the treatment by delivering the maximum dose to the target volume while limiting the dose to the critical structures [2], by precise determination of the dose distribution in tissues irradiated by the applicator.

We have developed a gauge to determine the three-dimensional (3-D) distribution of dose rate in water around ophthalmologic applicators. A miniature plastic $\phi 2 \times 2$ mm scintillator coupled to a photomultiplier tube (PMT) was installed on a mechanical arm which enables precise and reproducible 3-D measurements of the dose rate in water around the applicator to be achieved. Our gauge was calibrated for ^{106}Ru applicators, it can, however be easily adapted to measure 3-D dose rate distributions around ^{125}I ophthalmic applicators.

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Principle of operation

The principle of operation of the gauge is illustrated in Figs. 1 and 2. The investigated applicator is fixed inside a container filled with water. A scintillation probe employing a plastic $\phi 2 \times 2$ mm scintillator coupled to a PMT by a conical lightguide is used as the beta radiation detector. The container with the source can be moved reproducibly in horizontal and vertical directions. Additionally, the container can be rotated around its vertical axis, permitting dose rate to be measured at any point above the source area. A replaceable bronze $\phi 2 \times 23$ mm collimator can be attached to the scintillator to verify the uniformity of the source distribution over the surface of the applicator. The scintillation probe remains fixed during measurement. A fine micrometer screw is used to move the container with the source in horizontal and vertical directions with an accuracy of 0.01 mm. The range of horizontal and vertical movements are 25 and 50 mm, respectively, although vertical displacement over a range of 25 mm is adequate for the ^{106}Ru source (the 50 mm vertical range may be useful in the case of sources of more penetrating radiation). The overall accuracy of setting the radiation source with respect to the scintillator is within 0.1 mm or better. The output pulses from the PM, corresponding to the detected beta particles, after amplification and pulse discrimination, are fed to a microprocessor-controlled system where they are counted and processed. Pulses from the scintillation probe are formed to semi-Gaussian shape with a width of $1 \mu\text{s}$ at 10% of the pulse amplitude. A base line restorer maintains the base line level at zero value. An automatic gain control circuit is employed for on-line control of the scintillation probe gain. A light-emitting diode attached to the light-guide between the scintillator and the PMT photo-cathode, fed by a current pulse generator, produces reference light pulses. The amplitude of the light pulses is sensed by two pulse discriminators, E2 and E3, and the high voltage of the PM tube adjusted by a microprocessor (through a digital-analogue converter). The PMT high voltage is set to maintain the amplitude of the reference light pulses within at a midpoint between levels determined by the E2 and E3 settings.

Apart from measurements of the dose rate, the radiation activity of the source can also be evaluated with the described gauge. To measure the overall activity of the investigated source the scintillation probe is raised vertical-

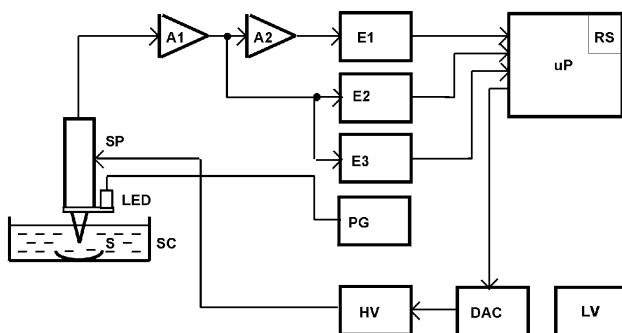


Fig. 1. Functional diagram of the gauge for measuring dose rate and ^{106}Ru activity distributions in an ophthalmological applicator. SC – movable water phantom; S – investigated radiation source (applicator); SP – scintillation probe; LED – light emitting diode; A1, A2 – pulse amplifiers; E1, E2, E3 – pulse discriminators; PG – pulse generator; HV – high voltage power supply; DAC – digital-analog converter; LV – low voltage power supply; uP – microprocessor; RS – serial port.

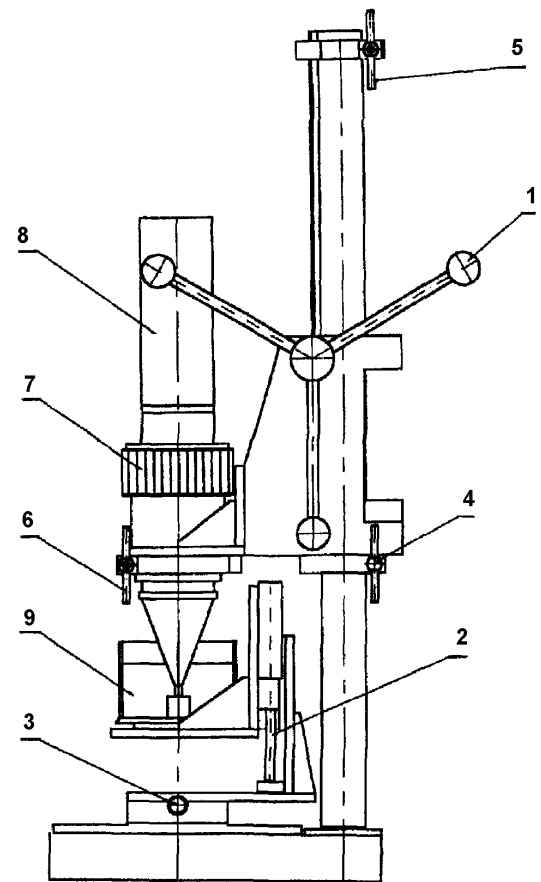


Fig. 2. Measuring head of the dose rate and activity gauge for ^{106}Ru sources. 1 – handle for moving the scintillation probe in the vertical direction; 2 – fine vertical movement screw; 3 – fine horizontal movement screw; 4 – lower limit of scintillation probe movement; 5 – upper limit of scintillation probe movement; 6 – scintillation probe lock; 7 – fine vertical adjustment of the scintillation probe; 8 – scintillation probe; 9 – movable container.

ly 250 mm above the source (from the lower to the upper fixed position of the probe), Fig. 2. At such source-detector distances the ratio of the distance to source diameter is no lower than 10 and the source can be considered as a point source, to an acceptable approximation.

The results of measurements obtained within a programmed counting time 20...120 s, are displayed and stored in the memory of the microprocessor together with their parameters and comments (date and time of measurement, co-ordinates of the scintillator with respect to the source, type of source, type of measurement) and are next transmitted by a RS232 serial port to an external PC computer. When processing the signal from the scintillation probe (pulse count rate), dead time correction is made. The gauge is equipped with an internal pulse counter facilitating the calibration of the gauge. Some auxiliary program functions (manual setting of the PMT high voltage, reference light emitting diode on/off) together with an internal pulse counter enable the operator to verify the correctness of gauge settings.

The electronic parameters and stability of the gauge

Calibration of the gauge was performed to relate the measured pulse-rate (at a fixed discrimination level) to the dose rate in water at a given point in the water phantom.

The relationship between the count rate and the measured dose rate is influenced by many factors, such as the pulse discrimination level, the counting channel dead time, spurious scintillation inside the light-guide due to radiation incident from the source and scattered radiation, the dependence of scintillation efficiency on radiation energy, etc. All measurements and calibrations performed in this work were carried out using two ophthalmic Ru-106 applicators of 12 mm curvature radius and diameters 15.5 mm or 20 mm, manufactured by Bebig GMBH, Germany, type CCA ($\phi 15.5$ mm) and CCB ($\phi 20$ mm), each of activity of about 4 MBq, after their application in eye brachytherapy.

To compensate for the dead-time effect in the measuring channel, dead-time compensation is made according to the relation $r=r_m/(1-\tau r_m)$ [2] where: r is the compensated count rate, and r_m – the measured count rate, τ – being the dead time of the measuring channel. Expected maximum count rate is approx. 150,000 pulse/s, and the dead time of the measuring channel is 1 μ s.

The effect of spurious scintillations induced in the light guide on the calibration procedure was verified by measuring the count rate with and without the scintillation detector in the scintillation probe. In both cases the scintillator housing touched the source surface in the centre of the source. A considerable increase of the count rate with respect to background was found (background approx. 1 p/s, the count rate due to scintillations in the light-guide: 280 p/s), at the selected discrimination level, however this was only 0.4% of the count rate measured in this position when the scintillator detector was in place. It was decided that a 0.4% relative increase of the induced number of scintillations in the light-guide can be neglected.

It is expected that the dose rate is proportional to the current of PMT generated by the light pulses in the scintillation detector. In the case of pulse mode operation of the PMT used for the determination of dose rate in the described gauge, the dose rate should also be proportional to the charge carried by the analogue pulses. To verify this expectation, analogue pulses from the output of pulse amplifier (A2 in Fig. 1) were fed to a rate meter, and the rate meter voltage (proportional to the current carried by analogue pulses) was measured at different points over the

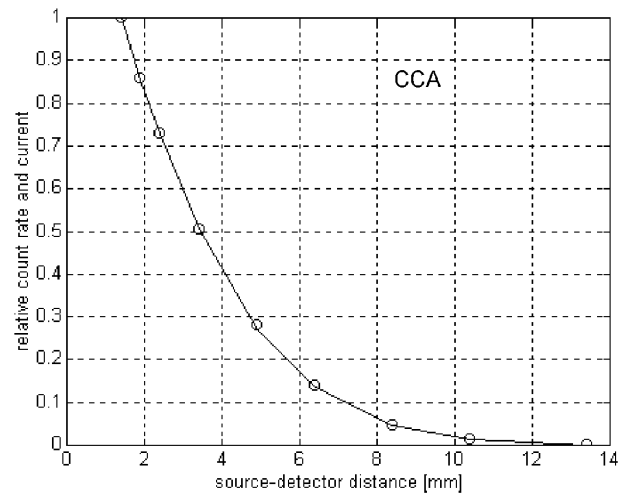


Fig. 3. Relative count rate (continuous line) and relative current (dots) against source-detector distance, normalized at 1.4 mm distance. The source-detector distance is the distance from the centre of the scintillator to the surface of the source along the central axis of the CCA source. Curves are normalized at 1.4 mm source-to-detector distance.

sources (not only along the central axis). Simultaneously, count rates at a fixed discrimination level were measured. An example of relative count rate and relative current (rate-meter output voltage) against the source-detector distance, are shown in Fig. 3. It was found from the measurements that the relative pulse count rate and relative current coincide to within 10.4 mm source-detector distance for the CCA source and up to within 8.4 mm for the CCB source with a deviation not larger than 5% in the worst case. One likely reason that at higher source-detector distances the relative count rate does not coincide with the relative current variation is a modification of the beta energy distribution, due to transport in the water phantom, as shown in Fig. 4. The difference in the shapes of the two spectra measured in water at source detector distances of 1.4 mm and 9.4 mm, is readily visible.

The gauge is equipped with an automatic gain control circuit of the PM tube which is activated automatically on switching the mains on. This solution eliminates any long-term PM tube gain variation and long-term temperature variation. The error due to the inaccuracy of setting the high voltage of PM tube, which can be set in 2 V steps (0.2%), results in a 0.4% error of the count rate. At expect-

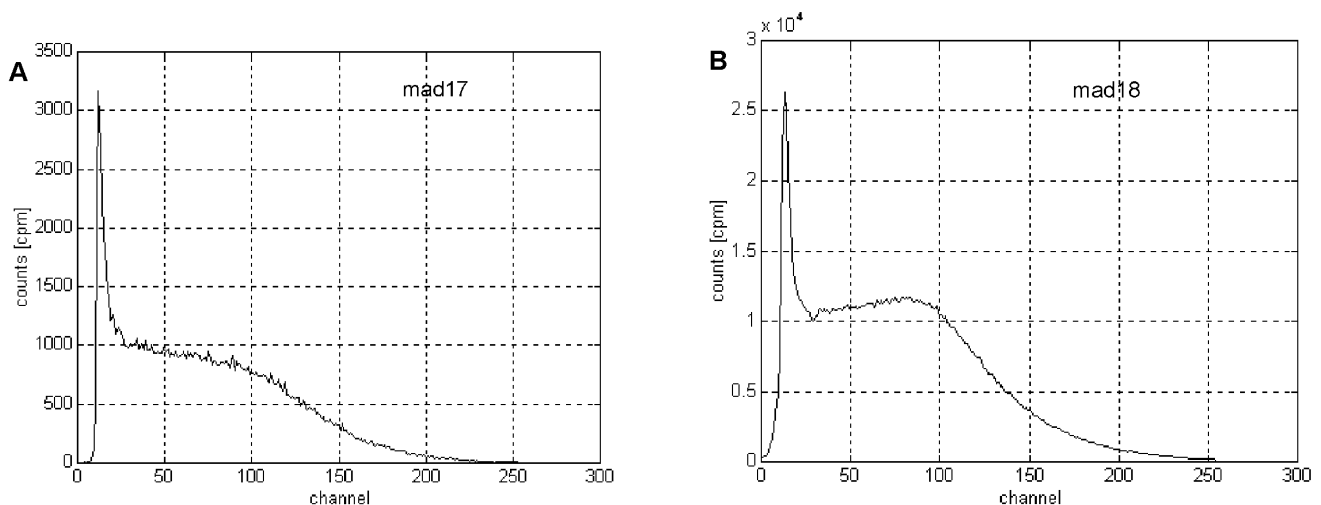


Fig. 4. Differential spectra of ¹⁰⁶Ru (CCA) measured in water at a distance of 9.4 mm (A) and 1.4 mm (B) from the centre of the scintillation detector to the source surface. The pulse discrimination level is set at channel no. 28.

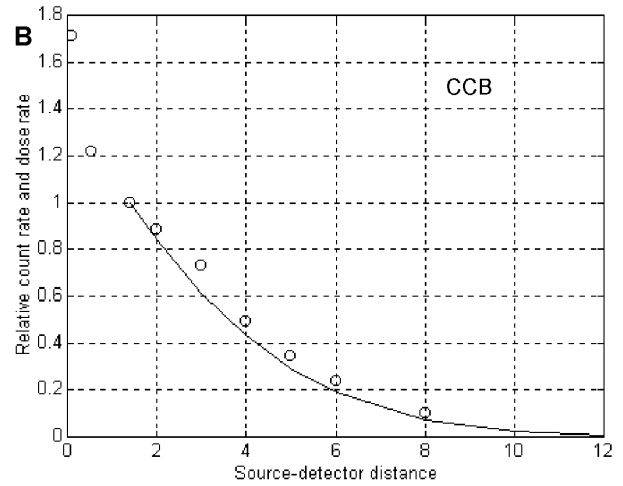
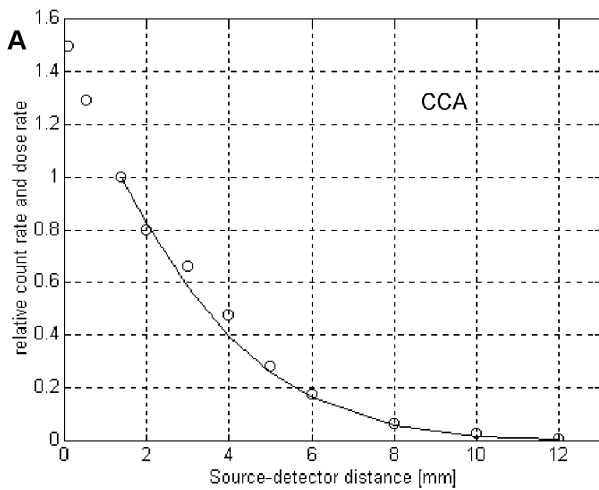


Fig. 5. Relative count rate (continuous line) and relative dose rate (dots) against source-detector distance in water, from $\phi 15.5$ mm (A) and $\phi 20$ mm (B) ^{106}Ru sources, normalized at a distance of 1.4 mm.

ed maximum count rate up to about 150,000 c/s the random error due to statistical fluctuations is well below 0.5%, depending on the source-detector distance and counting time. The count rate instability over 7 hours of operation of the gauge was found not to exceed 2% peak-to-peak.

Calibration

The gauge was calibrated using the recommended procedures and detectors [1]. Thin-layer sintered lithium fluoride (LiF:Mg,Ti) MTS-Ns thermoluminescence detectors (TLD), of 2 mm diameter and 0.1 mm thickness of the active layer, produced at the Institute of Nuclear Physics, Kraków, Poland were used in the measurements. The geometrical size of the detector corresponded to the size of scintillator and the thin active layer allowed us to perform measurements with high spatial resolution. The detectors were annealed at 400°C (1 h) + 100°C (2 h), sealed in a 0.05 mm-thick plastic foil (to protect the TL detectors against water), mounted on the measuring head of the gauge and irradiated in the water phantom at different source-to-detector distances. Prior to irradiation of the detectors, the pulse count rate was measured from the source used to irradiate the TL detectors. Separate irradiation time was set for each detector depending on its distance from the source, ensuring that the dose measured by detector at the each depth was approx. 0.5 Gy, i.e. within the linear response

range of LiF detectors. After irradiation, the protecting 0.05 mm foil was removed and the absorbed dose was read out, using a TLD reader.

The MTS-N detectors were calibrated in water in a therapeutic ^{60}Co beam at the Centre of Oncology, Kraków, Poland (the energy spectrum of secondary electrons from ^{60}Co is comparable with the slowing-down energy spectrum of ^{106}Ru electrons). The estimated total relative error of dose rate measurement is 8.1%. Results of measurements showing the relative count rate and the relative dose rate against source-detector distance, are shown in Fig. 5. The $\phi 2 \times 2$ mm scintillator is placed in a 0.4 mm-thick housing, and the minimum distance between the centre of the scintillator and the source is 1.4 mm. This source-to-detector distance was taken to normalize the dose rate and count rate curves. Examination of Fig. 5 shows that the relative dose rates measured with TLD detectors are, on the average, higher by 14% for the CCA source, and by 17% for the CCB source, than the relative count rate from the scintillation detector. The reason for this discrepancy is most likely the different thickness of the scintillation detector employed in the gauge (2 mm), compared with the thickness of TLD detectors used in the calibration measurements (0.1 mm).

To eliminate the systematic error and to match the indication of the dose rate to the count rate a coefficient k_z depending on count rate is introduced and the dose rate \dot{D} is determined from the relation:

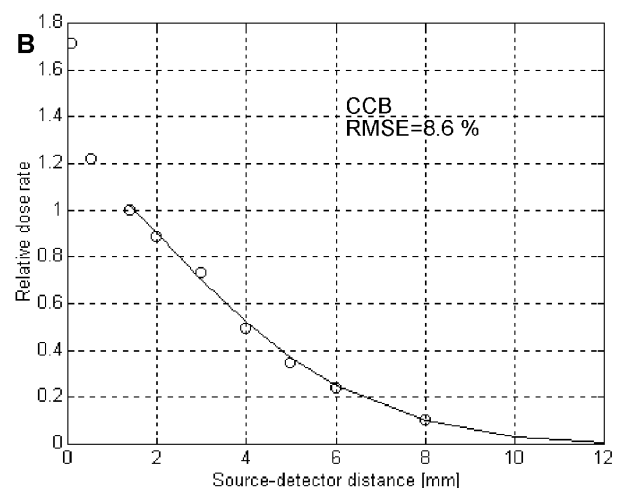
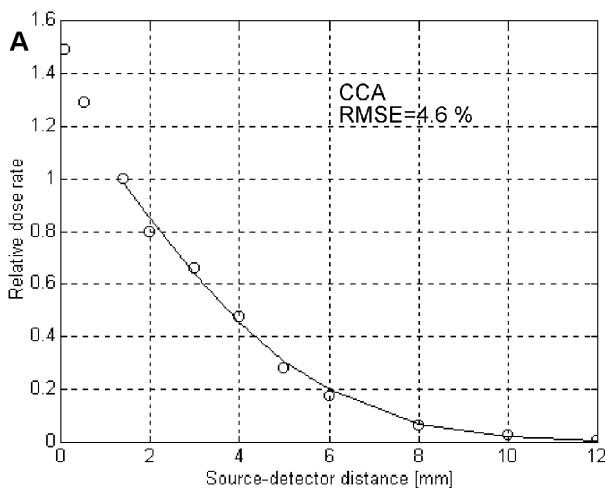


Fig. 6. Dose rate computed from the regression curve (continuous line), and measured (dots), for CCA (A) and CCB (B) ^{106}Ru ophtalmic sources.

$$(1) \quad D = D_0 k_z = n_0 k_d k_z$$

where: D_0 – dose rate at the distance of normalization = 1.4 mm; k_z is computed from polynomial fit of relative dose rate against relative count rate and is given by the relation $k_z = a_2 x^2 + a_1 x$, where a_2, a_1 – polynomial fit coefficients; $x = n_z/n_0$ – relative count rate at a distance z to the count rate at the normalization distance (1.4 mm); k_d – calibration coefficient (mGy/min)/(c/s) at the distance of normalization. The relative dose rate and relative count rate against source-detector distance computed from Equ. (1), normalized at a source-detector distance of 1.4 mm, is shown in Fig. 6. The dose rate at 1 mm distance from the source with respect to the dose rate at 1.4 mm is higher by 14% for the $\phi 15.5$ mm source and 16% higher for the $\phi 20$ mm source, respectively. The k_d coefficient for the $\phi 15.5$ mm and $\phi 20$ mm source is the same and equal to 5.8×10^{-4} (mGy/min)/(c/s), but the shapes of the regression curves are somewhat different. The Root Mean Square Error of dose rate against count rate (distance) in Fig. 6 does not exceed 8.6% in the worst case, and the mean reading of dose rate in respect to count rate does not exceed +2% for measuring points up to 12 mm source-detector distance. Indications given in [1] require that the dose rate reading does not differ more than 10% at a distance of 5 mm from reference curve. This requirement is fulfilled with better accuracy.

Conclusions

The $\phi 2 \times 2$ mm scintillation plastic detector employed in the gauge, coupled to a photomultiplier tube (PMT) operating in pulse mode, features high detection efficiency of beta radiation, high pulse count rate, resulting in a low 0.5% random error due to statistics of detecting ionizing radiation. The PMT, the most unstable component of the measuring channel, is controlled by an automatic gain con-

trol circuit, activated as the mains supply is switched on, prior to measurement. The long-term gain variation due to fatigue of the PMT or ambient temperature variation is thus compensated for. The count rate error due to the inaccurate setting of the high voltage of the PMT is 0.4%, and the instability error over 7 h of continuous operation of the gauge does not exceed 2%, peak-to-peak.

The ratio of the pulse count rate to the current carried by analogue pulses is constant at source-detector distances in the range from 1.4 mm (the probe touching the detector surface) to 8.4 mm, with an accuracy better than 5%, in the worst case. Above that distance the ratio is not constant and may change up to 30%. On the average, the relative count rate (and charge carried by analogue scintillation pulses) in the range up to 8.4 mm of source-detector distance, differs from the relative dose rate against the distance by 14–17%. To compensate for these two effects and to match the count rate curve to the dose rate curve a non-linear regression coefficient is used to evaluate the dose rate.

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