Purpose. The purpose of this study was to calibrate different gamma cameras in the framework of a multicentre research for lesion dosimetry in 223Ra therapy of bone metastases. Equipment of several manufacturers and different models were available. Moreover, differences in gamma camera calibration procedures into activity quantification were explored.

Methods. Eleven gamma cameras of different crystal thickness (3/8- and 5/8-inches) were used, acquiring planar static images with double-peak windows (at 82 and 154 keV, 20% wide) and a MEGP collimator in all cases. Sensitivity was measured in air, varying the activity, the source-detector distance and the diameter of the source. Transmission curves were measured, calculating the parameters used for attenuation/scatter correction with the pseudo-extrapolation number method, and assessing their variations with the source size.

Results. The values of the calibration factors (considering the geometric mean of both detector sensitivities) ranged from 41 to 114 cps/MBq. For the smallest source (diameter of 3.5 cm), the calibration factor decrease ranged from 30% to 4%, highlighting the importance of partial volume effect according to the equipment involved. Partial volume effects were negligible for an object area of 960 mm². The sensitivity was nearly constant varying activity and distance, and its variation with the source-detector distance, with respect to the 15 cm-value, reached 10% (in absolute value) in the range of 5–30 cm. Fixing the distance between the two heads, the calibration factor remains within the distance from the midline was within 3.6%. Examining the results obtained with six gamma cameras, appreciable variation of the transmission curves with the source size and experimental setup were observed, leading to activity quantification errors up to 20%.

Conclusion. The calibration protocol that should be regularly implemented for Ra22 acquisitions requires sensitivity and transmission curve measurements varying the source size, and performing a careful procedure of standardisation. The results of this study represent a compendium useful for dosimetric purposes.

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[1007] Treatment planning in proton therapy
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Purpose. To present the most important physics aspects of proton therapy (PT) treatment planning for scanning beams and compare them with photon-based radiotherapy.

Methods. The most important differences of protontherapy treatment planning with pencil beam scanning (PBS) with respect to what is considered now the norm in radiotherapy with photons (XRT) are the following:

1) The dosimetric inaccuracies in the map of proton stopping power (PSP) estimated via CT imaging are larger than similar error in estimating electron density for photons. This is the reason behind the development of more refined methods (e.g. dual energy CT or proton CT) and the current impossibility of MR-based protontherapy;
2) The pristine beam is in principle simpler to model than in photons, as no beam modifiers are used to shape the beam, with one significant exception: the preabsorber (aka “range shifter”);
3) Up until recently, dose calculation was performed with pencil beam-based algorithms. Now a transition is happening towards Monte Carlo-based algorithms, however for reasons that are different than in XRT;
4) In XRT geometrical uncertainties are handled during planning with ad hoc volume such as the planning target volume (PTV). In PT this approach is hardly satisfactory, albeit still common, and most advanced planning solutions are in fact based on PTV-less so called “robust optimization”, where the effect of range and setup uncertainties is essentially part of the cost function. This leaves open the issue of dose reporting, which is now approached in different ways at different PT centers;
5) PT treatment planning is performed assuming a constant 1.1 relative biological effectiveness (RBE) of protons with respect to XRT. This is an approximation, and the need of variable RBE in proton planning is one of the “hot topics” at the moment;
6) Adaptive therapy, and as a consequence adaptive treatment planning, is an area of development that has to be explored further. PT may be a field where the concept of “online adaptive” may have a clinical impact, but the times are still premature for a large scale implementation of these techniques.

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[1008] Systematic evaluation of lung-tumor motion using four-dimensional computed tomography
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Purpose. To establish a suitable routine for handling of internal patient motions during radiotherapy delivery it is valuable to have information on the actual motions in a large group of patients. This presentation focus on a study (Sarudis et al., Acta Oncologica 2016; 56) that provides information on the lung-tumor motions due to breathing for 126 patients treated with stereotactic body radiotherapy.

Methods. Four-dimensional computed tomography images were reviewed. The tumor motion was determined by the center-of-mass shift between the breathing phases containing the largest positional differences. Patients were evaluated separately in subgroups depending on tumor diameter (Ø <2.0 cm, 2.0 < Ø < 5.0 cm, Ø > 5.0 cm) and location within the lung (upper, middle, lower lobe). The motion pattern in each direction (interior-superior, left-right, and anterior-posterior) were analyzed for tumors moving >5 mm and sinusoidal trigonometric functions were fitted using the least mean square method. Mann-Whitney statistical tests were used for statistical analyses.

Results. Tumor volumes were between 1.6 and 52.3 cm³. The mean and maximum amplitudes of the tumor motions were 6.9 and 53.0 mm (inferior-superior), 1.5 and 11.0 mm (left-right), and 2.5 and 9.0 mm (anterior-posterior). 95% of the tumors moved ≤20 mm, ≤3 mm, and ≤6 mm in the inferior-superior, left-right and anterior-posterior directions respectively. The observed motions showed no correlation with tumor size or location except for motions in the inferior-superior direction where the motion amplitude significantly increased for tumors located in the middle compared to the upper as well as in the lower compared to the middle part of the lung. The motion pattern of a tumor was always best described using a squared trigonometric function: A \cdot \sin^2(\varphi - \beta)$, where A = maximum amplitude, t = time of measurement, T = total time of the breathing cycle and B is a constant used to synchronize the starting point of the breathing cycle.