

ORIGINAL ARTICLE

An overview of volumetric imaging technologies and their quality assurance for IGRT

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Abstract

Image-guided radiation therapy (IGRT) aims at frequent imaging in the treatment room during a course of radiotherapy, with decisions made on the basis of this information. The concept is not new, but recent developments and clinical implementations of IGRT drastically improved the quality of radiotherapy and broadened its possibilities as well as its indications. In general IGRT solutions can be classified in planar imaging, volumetric imaging using ionising radiation (kV- and MV- based CT) or non-radiographic techniques. This review will focus on volumetric imaging techniques applying ionising radiation with some comments on Quality Assurance (QA) specific for clinical implementation. By far the most important advantage of volumetric IGRT solutions is the ability to visualize soft tissue prior to treatment and defining the spatial relationship between target and organs at risk. A major challenge is imaging during treatment delivery. As some of these IGRT systems consist of peripheral equipment and others present fully integrated solutions, the QA requirements will differ considerably. It should be noted for instance that some systems correct for mechanical instabilities in the image reconstruction process whereas others aim at optimal mechanical stability, and the coincidence of imaging and treatment isocentre needs special attention. Some of the solutions that will be covered in detail are: (a) A dedicated CT-scanner inside the treatment room. (b) Peripheral systems mounted to the gantry of the treatment machine to acquire cone beam volumetric CT data (CBCT). Both kV-based solutions and MV-based solutions using EPIDs will be covered. (c) Integrated systems designed for both IGRT and treatment delivery. This overview will explain some of the technical features and clinical implementations of these technologies as well as providing an insight in the limitations and QA procedures required for each specific solution.

Image-guided radiation therapy (IGRT) aims at acquiring anatomical information of the patient in the treatment room to make decisions based on this information and hence improving the quality of the treatment. Due to the increasing computing power, the introduction of automated tools for image registration and quantification of set-up errors became a reality and this combined with a good integration in the workflow promoted IGRT to be generally adopted in mainstream clinical routine. Many reviews have been written on the subject the last years [1–3]. Volumetric imaging and in particular CT offers an attractive solution as it provides, non-invasively, accurate geometric information of the tumour as well as its relationship to surrounding sensitive tissue. There are generally two main applications for using in-room acquired volumetric information: a prospective or (on-line) correction

protocol where the image data set is registered with a reference data set to guide the patient set-up prior to treatment, or a retrospective approach. The latter can either be generating statistical data of consecutive set-up sessions to help determine an optimal PTV/PRV (planning target volume and planning organ at risk volume) margin, or reconstruct the dose received by the patient using the daily CT-image data sets to evaluate the cumulated dose distributions from several sessions and define strategies for adaptive radiotherapy (e.g. re-optimizing the treatment to cope with uncertainties in patient set-up, changes in target volume or anatomical variations). Much effort in this field of research focuses on improving image quality, and as with diagnostic imaging, non-radiographic volumetric imaging techniques show much potential. One of the most obvious solutions (although not straightforward in

application) in that respect would be to integrate magnetic resonance imaging for IGRT and research in this field looks promising [4]. Most volumetric IGRT solutions produce a static volumetric image data set and can therefore only account for inter-fractional variability in patient set-up, target localization or anatomy. This technology is still an area of continuing development and techniques allowing cine-mode approaches [5] or respiratory correlated cone-beam CT (CBCT) [6–8] are currently being investigated. These techniques can help to assess if the PTV margin suffices for adequate tumour coverage or evaluation of motion management. Future developments include techniques that will enable real-time dose calculation based on these images-of-the-day. Volumetric imaging data acquired at the treatment machine offers the potential of accurately estimating the actual dose delivered for assessment in patient follow-up or adapting the treatment plan to adjust for deviations of the ideal dose distribution. In general one can say that (at the time of writing) planar imaging techniques offer the possibility for real-time assessment of target localization during treatment provided fiducials are present to identify the target's position (i.e. surrogates such as bony structures or implanted markers), whereas volumetric imaging techniques offer volumetric data of both target volume and surrounding structures without the need of any surrogate but compromising the real-time aspect. In this review we will only evaluate the different systems' performances to determine the relative displacement of the target volume with respect to its planned location, based on information obtained from image registration tools.

As always, the idea is not new [3] and already in the 1980's Simpson *et al.* [9] explored the idea of using the treatment beam to generate single slice tomograms with one gantry rotation of the linac. Currently a number of solutions adopting CT image acquisition have been introduced in clinical routine including the use of a dedicated CT scanner in the room (high-end diagnostic CT scanner [10–12] or C-arm solution [13]), on-board devices for cone-beam CT mounted to the treatment machine (kV [14–17] or MV [18,19]) or fully integrated CT systems such as the helical TomoTherapy approach [20]. The imaging aspects of these technologies have been explored and reported in literature, and the reader is kindly referred to these publications for in-depth information. However, as these systems are designed to (re)position the patient before/during treatment the geometric accuracy is most critical. When integrating volumetric IGRT data for targeting treatment beams an accurate relationship between the image reconstruction space and the treatment isocentre must be established and maintained in a clinical set-up. In this

paper some of the more common approaches for volumetric IGRT will be reviewed with special emphasis to geometric accuracy and QA.

Dedicated in-room CT-scanners

The scanner may be positioned over the treatment couch using rails and/or the treatment couch may be used to transport the patient into the bore of the CT gantry, or even the use of a C-arm CT system can be considered. An illustration of the in-room CT system is given by Court *et al.* [11] reporting mechanical precision and alignment uncertainties for this integrated CT/linac system. The system described integrates a high-speed CT scanner on rails and a linac. The couch base can be rotated to position the patient for either treatment or scanning, without having to move the patient from the treatment table to the CT couch. Concerning geometric accuracy the following sources of uncertainty have been identified: (1) the patient couch position on the linac side after a rotation, (2) the patient couch position on the CT side after a rotation, (3) the patient couch position as indicated by the digital read-out, (4) the difference in couch sag between CT and linac positions, (5) the precision of the CT coordinates, (6) the identification of fiducial markers from CT images, (7) the alignment of contours with structures in the CT images, and (8) the alignment of set-up lasers. The largest single uncertainty (1 SD) was found in the couch position on the CT side after a rotation (0.5 mm in the lateral direction) and the alignment of contours with the CT images (0.4 mm in the cranio-caudal direction). All other sources of uncertainty were less than 0.3 mm (1 SD). The major advantage of this approach is the availability of a high-end CT scanner with optimal image quality, which was confirmed in a recent publication by Stuetzel *et al.* [21]. The combination of a diagnostic CT-scanner (Siemens Somatom Emotion®) on rails with a gantry-based linac (Siemens Primus®), proved to yield superior image quality compared to some on-board imaging solutions when considering the soft tissue contrast achievable with low imaging doses.

An alternative approach has been investigated by Sorensen *et al.* [13] introducing a flat panel mobile C-arm, capable of kV CBCT, into the treatment room. A commercial optical tracking system (ExacTrac®, BrainLAB) has been introduced to define the relationship between the C-arm image co-ordinates and the treatment machine isocentre, so as to obtain the appropriate reference frame for the reconstructed images. The system can rotate in synchrony with the linac allowing image acquisition in treatment position avoiding collisions between both devices. A default

infrared (IR) isocentre calibration phantom known from the commercial ExacTrac[®] device is introduced to define the linac isocentre within the IR tracking system reference frame. This enables accurate location of any IR-reflecting object in treatment room coordinates, which subsequently is used to track the C-arm within the treatment room. Localizing reflective markers on a phantom with both the infrared tracking device and transforming the reconstructed CT-images to calculate room coordinates, the authors reported a mean absolute difference of $1.4 \text{ mm} \pm 0.5 \text{ mm}$. As the calibration procedure is based on room-laser alignment in defining the room coordinate system, a fundamental assumption of this approach is that the room lasers are “perfectly” aligned with the linac isocentre.

On-board kV CBCT

Jaffray et al. [14] were among the first to explore the kV CBCT concept integrated with existing treatment machines (‘on-board kV CBCT’ will be used in the current paper to coin this particular approach, and although the term has been marketed by a particular vendor it is not limited to a particular commercial system) and several manufacturers have adopted this approach in recent releases of their equipment. The concept is based on integrating a kV x-ray source and a large-area flat panel detector on a standard linac allowing fluoroscopy, radiography, and volumetric kV CBCT. The kV imaging chain can be mounted orthogonal to the treatment beam (Elekta, Varian) or in-line with the treatment beam (Siemens). The former has the advantage in that stereoscopic planar imaging is possible with one gantry position by applying hybrid kV and MV imaging provided an electronic portal imaging device (EPID) is also installed. The inline approach has the advantage in that the kV imaging axis coincides with the axis of the treatment beam, providing on-line image data in the beam axis. The kV CBCT allows a volumetric CT image to be reconstructed from data collected during a single gantry rotation. Because the gantry rotation of a linac is much slower than a CT ring gantry, flat-panel detectors are introduced to acquire so-called cone-beam CT or volume CT imaging. The design of currently available on-board kV CBCT systems is, however not optimal. Its quality is adversely influenced by many factors, such as scatter, beam hardening, mechanical instability and intra-scanning organ motion.

The three most common commercial solutions present different features in basic design that influence the approach for QA. However, all systems share the principle that the kV imaging isocentre and the MV treatment isocentre are independent and

might not necessarily coincide exactly. The QA for geometric accuracy needs to account for this feature. Moreover, due to tube sag during gantry rotation and projection angle uncertainties ‘systematic’ geometrical uncertainties are introduced and no perfect geometry can be achieved for the image acquisition. The Elekta and Siemens approaches recognize the existence of mechanical flex and in assuming that this is reproducible and stable in time, a correction is introduced in the image reconstruction algorithm. The QA procedure, therefore, primarily focuses on verifying the reproducibility of this mechanical flex [15,16]. The Varian approach, on the other hand claims a strong rigidity of the mechanical arms assuming the influence of the remaining wobble is negligible. Note that Siemens offers both a kV CBCT as well as an MV CBCT solution.

Jaffray et al. have proposed the kV CBCT system [14] that has recently been commercialised by Elekta. The x-ray tube – detector axis is orthogonal to the treatment beam. A conventional x-ray tube has been mounted on a retractable arm that extends from the accelerator’s drum structure and a $41 \times 41 \text{ cm}^2$ flat-panel x-ray detector is mounted opposite the kV-tube at a nominal detector-to-focal spot distance of 155 cm. These investigators identified two major geometric non-idealities with the first prototypes (variations in the angular velocity of the gantry of a factor 2 through rotations over 360° , and variability in the geometric relationship between the kV focal spot and the flat-panel detector attributed to flex in the detector motion) that in the end have been taken into consideration during reconstruction by adjusting the back-projection to be consistent with the geometry of acquisition. Based on phantom studies Jaffray et al. [14] illustrated the full volumetric nature of the cone-beam CT data, showing excellent spatial resolution in all 3 dimensions (as opposed to conventional fan-beam CT where the cranio-caudal resolution depends on the slice thickness and pitch). The system provided sub-millimetre spatial resolution (approximately 0.7 mm full-width at half maximum of the line spread function) and a lowest readily detectable contrast at 47 Hounsfield Units (HU). Sharpe et al. [15] performed phantom measurements to assess the alignment of the centre of kV CBCT reconstruction to MV radiation isocentre to image and localize a 1 cm diameter steel ball bearing (BB). Data from 21 treatment sessions over a 3 month period acquired on a linac in routine clinical use, demonstrated that the system was able to relocate the object (the kV CBCT data was registered to the planning data to assess couch corrections) within less than 1 mm of the prescribed location. The mechanical isocentre of the kV system was found to be within $0.5 \text{ mm} \pm 0.5 \text{ mm}$ of MV radiation

isocentre. The remaining set-up accuracy depended on the mechanical precision of the different components of the delivery system. These authors also presented a simple method for geometric calibration using the above mentioned BB. In a first phase the BB (initially placed at isocentre with the help of room-lasers) is aligned with respect to the treatment isocentre based on orthogonal MV portal images acquired with an EPID. Consequently, look-up tables are generated by calculating the centroid of the BB in each projection of the kV CBCT acquisition as a function of gantry angle. These look-up tables are stored for subsequent use and employed during the reconstruction process to relate each projection to the reconstructed voxel array, thereby assuring that the centre of the reconstructed volume is coincident with the estimated treatment isocentre (based on MV-imaging). Again, this procedure assumes that the mechanical characteristics of the system are systematic and reproducible. A similar approach has been reported for the Siemens solution by Oelfke et al. [16]. Yoo et al. [17] reported a multi-institutional verification of the Varian volumetric IGRT solution. Again based on BB measurements, a mechanical isocentre accuracy of less than $1.5 \text{ mm} \pm 1.0 \text{ mm}$ was reported for 60 measurements spanning a time period of 6 months. No specific measurements on geometric accuracy of the kV CBCT images have been reported. Important to note with the Varian approach, is that the kV source can move towards and away from the isocentre as well as in the cranio-caudal direction of the patient. The detector has an additional freedom of motion parallel to the treatment axis.

MV CBCT

The use of EPIDs available on a conventional linac to produce megavolt cone beam CT image data (MV CBCT) has both inherent advantages as well as disadvantages. As with EPIDs, contrast is poorer compared to diagnostic x-ray quality, on the other hand high-Z artefacts are not present. The latter not only reduces imaging artefacts caused by (dental) prosthesis or even bone but also improves the unique identification of implanted radio-opaque markers. Application of the EPID to generate MV CBCT has the advantage that no additional hardware is required. And finally, alignment of target and treatment beam is straightforward as the actual treatment beam is used to generate the images. Nevertheless, similar to the kV approach, mechanical instabilities in the detector assembly will still influence the image reconstruction. Feasibility studies on MV CT scanning had been performed in the 80's and were typically based on a single slice tomogram per gantry

rotation [9]. A major problem with these approaches was accurate table indentation. Nakagawa et al. proposed to use a pre-treatment MV CT slice to verify the patient set-up for stereotactic radiosurgery of the lung [22]. To overcome the problem of table indentation Mosleh-Shirazi et al. [23] reported a feasibility study on 3D MV CBCT using a scintillation detector – CCD camera based EPID on the linac, with the image frame acquisition synchronized with the radiation pulses. A first prototype required approximately 40 cGy and 2h reconstruction time on a Sun SPARC 2 to obtain a density resolution of 2% and spatial resolution of 2.5 mm. Ford et al. investigated the use of gated image acquisition to reduce motion artefacts and defining a limited region of interest [18], their system required 2.5 MU/projection and 100 projections (approx 7 min.) to yield 2% contrast resolution and 2 mm spatial resolution. A major concern in MV CBCT is the extra-target dose introduced by the target localization process due to the challenge posed by the poor detection efficiency of x-ray detectors in the MV energy range. This low efficiency results in poor signal-to-noise performance for clinical acceptable doses [14,21]. Current “low dose” solutions are possible with frame acquisitions during beam-off and a trigger mode yielding 0.08 MU/image frame or 15 MU in total for a volumetric MV CBCT [19]. These investigators showed the possibility of using a standard linac (Siemens) with stable low dose rate and an EPID to obtain clinically useful images. An interesting feature of MVCT is the linear relationship between electron density and megavoltage Hounsfield Units (HU) due to the almost dominance of Compton scatter as the attenuation mechanism for clinical megavolt beams (4–6 MV) for the tissue materials encountered in clinic.

MV CT

A fan beam MV CT approach is presented with helical tomotherapy, which is a 2-in-1 concept of a linac with a helical CT scanner. The ring-based geometry provides the high mechanical stability known from diagnostic CT scanners. This system uses a fan beam to acquire an MVCT of the patient prior to and potentially even during treatment [20]. For CT mode the accelerator is de-tuned from 6 to 3.5 MV and the pulse repetition frequency is decreased to keep the patient dose below 3 cGy. An arc-shaped Xe-detector array with 738 channels (of which 540 channels are used for reconstruction) is used for image acquisition. For treatment a dedicated binary MLC is used to modulate the fan beam to provide rotational IMRT, not unlike the add-on device for sequential tomotherapy (MIMiC,

NOMOS, Sewickly, PA) [24]. The beam rotation is synchronized with continuous longitudinal movement of the couch through the bore of the gantry, performing a helical beam pattern. When operating as a helical MVCT system, the leaves are fully retracted to the open state. Three modes of image acquisition (coarse, normal, fine respectively) obtained by different pitches are available resulting in image reconstructions with inter-slice distances of 6, 4 and 2 mm. The on-board CT option offers a number of verification processes: (a) The MVCT scan can be registered with the planning CT scan for automated target localization and positioning prior to treatment (currently only rigid registration is supported without the possibility of defining regions of interest). Verification of the automated registration routine on an anthropomorphic phantom showed correct translations and rotations to an accuracy of less than 1 mm or 1 degree [25]. The set-up correction (involving rotations and translations) can be implemented either by moving the patient or, in principle, by modifying the IMRT delivery to account for the patient's actual geometric offset. (b) The CT detector system can be operated during the treatment to compare the detector signal with the expected signal and as such detect deviations, or alternatively, to reconstruct the dose delivered to the patient from exit dose measurements. The energy fluence distribution and the CT representation can be used to compute the actual dose distribution in the patient. This reconstructed dose distribution represents the dose the patient actually received, and it may be superimposed on the CT representation just obtained to realize a new form of in vivo dosimetry. This last option still is work in progress. Again, the unit applies the actual treatment beam for image acquisition and most of the commissioning and QA procedures of the treatment device incorporate the use of the imaging detector array. As such the registration of image space to treatment space presents an integral part of the system's QA (e.g. alignment of ionisation chamber and central axis for absolute dosimetry is accomplished by MVCT imaging). Some of the basic QA procedures such as system synchrony (gantry-couch-collimator), system geometry, field size, field centring and alignment, and isocentre constancy have been described by Fenwick et al. [26].

Image registration tools

The advantages of having volumetric patient data at the time of treatment will not be fulfilled without image analysis software. The spatial relationship between images is established by a transformation

model that is optimized through a registration algorithm. Using rigid body transformations, set-up corrections involving translations and rotations can be implemented either by moving the patient or, in principle, by modifying the treatment delivery to account for the patient's actual geometric offset. As these so-called registration algorithms will be used for patient localization and setup, not only the translation from imaging coordinate reference system to treatment coordinate reference system, but also the reliability of the positional correction parameters requires careful verification. These registration results may be influenced by the imaging acquisition settings (i.e. image quality: spatial resolution, contrast, noise, etc.), the available anatomical information (e.g. scanned region), as well as settings of the registration algorithm itself (most software packets offer user-defined parameters such as regions of interest, soft tissue versus bone registration, etc.). A complicating factor is that the patient representations in the original planning CT and the CT-of-the-day may not be the same. Rigid transformations currently predominate but are challenged by these non-linear deformations associated with treatment response, weight change, variation of organ position and volume between examinations, and more frequently, differences in patient pose. In a first phase this problem can be solved by determining regions of interest to optimize the registration process (if possible with appointing different weights of importance to different anatomical regions [M. van Herk, ESTRO IGRT course, 2006]), or to apply a deformable registration [27]. The latter will be required to create cumulative dose distributions and adapting the treatment plan to take into account deficiencies of previous treatment sessions. Once the mapping from one image set to the other is established, the dose distribution calculated in one set can be mapped and added to the other using the same transformation. Most verification studies of registration algorithms for IGRT have been performed on particular systems and no general comparison or QA-tools are available. In general, guidelines for quality assurance of image registration tools in stereotactic radiosurgery and treatment planning are appropriate for IGRT. A typical example are the recommendations of the International Atomic Energy Agency (IAEA) for the assessment of the technical principles, possible bias to a particular modality, constraints imposed on image acquisition, the degree and behaviour of automation, the registration model (rigid or deformable) and dependence on image acquisition parameters [28]. Recently, the American Association of Physicists in Medicine (AAPM) formed task group 132 to review techniques for image registration, identify issues related to

clinical implementation, assess accuracy, and discuss acceptance and QA. Usually, some kind of visual verification tool (split screen displays, fast toggling between image sets or colour coding) will help the user to assess the quality of registration. However, the accuracy of the calculated correction parameters needs to be verified on phantom studies. Typically, phantoms with embedded radio-opaque markers can be applied for these image localization and position correction tests. One such example has been provided by Kashani *et al.* [29]. These investigators introduced a deformable phantom, embedded with small identifiable reference marks, to apply a known deformation to a simple geometry and quantitatively evaluate the outcome of the registration algorithm by comparing measured and estimated location of a series of points.

Volumetric acquired in-room image data for dose calculation

Volumetric patient data acquired on the treatment machine offers the potential advantage of performing dose calculation on these 'images-of-the-day' with the patient in treatment position to assess the actual delivered treatment dose. With the use of appropriate (non-rigid) registration tools to create cumulative dose distributions, this feature can be used as a verification tool or even allow adapting the treatment plan to adjust for deviations from the ideal plan. Compared to diagnostic images the CT-data acquired at treatment machines usually show enhanced noise and an increased level of artefacts mostly related to scatter radiation. The determination of accurate Hounsfield Units (HU) and the respective calibration to electron densities needs careful investigation if dose calculations are to be performed on these images. An additional challenge with volumetric image data acquired on a treatment machine is the limited Field of View (FOV). The absence of the complete anatomy of the patient hampers dose calculation. The Elekta, Varian and Siemens solutions respectively offer an axial FOV with 25.6 cm, 24.0 cm and 27.0 cm diameter, and 25.6 cm, 15.0 cm (can be extended to 17 cm by adapting the reconstruction algorithm) and 27.0 cm cranio-caudal dimension [15–17]. Most CBCT approaches try to overcome this problem by laterally shifting the detector off its central position. For Elekta this yields an increased diameter to 40.0 cm, Varian 45.0 cm and Siemens 48.0 cm. This mechanical procedure, however, requires an adaptation of the reconstruction algorithm to account for this new set-up of the imaging hardware. The cranio-caudal dimension of the scan area however, is fixed, or as is the case with helical tomotherapy often limited due to time

restrictions. The latter solved this problem by creating merged images in that the missing data (cranial and caudal from the scanned region) will be completed with the diagnostic CT-data acquired for initial treatment planning. The intrinsic MVCT FOV diameter in TomoTherapy is 40.0 cm. The electron densities in the merged image set have been obtained with the appropriate conversion data for accurate dose calculation. Several groups are investigating in this field of research and solutions will become commercially available soon, allowing dose calculation on the CT images-of-the-day. Again, as this is a fast developing field of research, no general conclusions can be drawn, and it is up to the user to investigate the recalculation technique and to determine its uncertainties (with special emphasis to reliability and stability of electron densities).

Volumetric IGRT and patient dose

As treatment outcomes will be improved, the concept of concomitant patient dose due to IGRT also needs to be revisited. Even with the development of dose saving acquisition modes, daily use of EPID will result in cumulative extra doses of about 1–2% for dose prescriptions of 70–80 Gy. It has to be noted that most IGRT approaches make use of imaging modalities that range from planar imaging to fluoroscopy to CT-based solutions that follow regimes as simple as acquiring single set-up images or as complex as assessment of intra-fraction tumour tracking. Radio-graphic IGRT solutions might have an important impact on extra patient dose both concentrated at the skin in case of planar kV x-ray imaging or distributed throughout the anatomical volume of interest when CT-based imaging is introduced. MV-based images, in principle can be incorporated in the dose calculation as the beam characteristics and biological effects have been modelled previously. The latter is not straightforward for kV-based imaging modalities. Patient dose is directly related to image quality and the latter needs to be optimized in function of the clinical requirements. For patient set-up in some cases, especially with automated registration algorithms, excellent soft tissue contrast may not be necessary and image quality may be compromised in favour of reducing patient dose (e.g. for positioning strategies based on bony structures as in many cranial or head-and-neck applications). An excellent review on this topic is provided by the task group 75 of the AAPM on the management of image dose during IGRT [30]. Indeed, the wide variety of modalities and the combination of added patient dose due to increased imaging in the preparation of treatment as well as changes in extra-target dose from recent developments in treatment delivery (e.g. IMRT and

particle beams) make it difficult to synthesize a complete picture of the patient's exposure, and as always the challenge will be to find a good balance between the associated risk for the patient and the expected benefit.

Conclusions

With the increasing availability of imaging and dosimetric information during a treatment course, the original treatment plan can be altered to tailor the dose distribution or control dose delivery in the presence of patient geometry variations or delivery errors. Volumetric IGRT techniques represent an important tool to accomplish this goal both as patient set-up tool as well as for dose calculation on images-of-the-day allowing adaptation of the treatment plan based on this data. Needless to say that this field of research is still in development and many possibilities will require careful validation. On the other hand, the tools are available to date, and as always the major challenge will be to develop strategies for optimal application of this technology in a clinical environment and recognizing both its possibilities as well as its limitations.

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