A simple method for evaluating stereotactic accuracy of Magnetic Resonance Imaging and Computed Tomography Imaging in Frame Based Radiosurgery*

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Abstract

Purpose

To design a simple and reproducible method with minimum uncertainty for evaluating the stereotactic accuracy of Computed Tomography imaging and Magnetic Reso-nance Imaging and to compare the results.

Methods

3D phantom has been designed with 1 mm diameter targets in eight sectors for stereotactic MR imaging and CT imaging. The phantom was connected to Leksell stereotactic frame and the images obtained are exported to treatment planning system. Leksell stereotactic coordinates (X, Y, Z) of the holes are determined in each slice and mean maximum errors were calculated along with 3D vector distances from center of stereotactic coordinate system (100,100,100) to five known Targets. The target volumes are calculated independently for each image data sets.

Results

The mean of maximum absolute error estimated for MR images from the Siemens Magnetom vision MR unit were 0.46 mm (X-Axis), 1.66 mm (Y-Axis) and 2.11 mm (Z-Axis). Mean of absolute maximum error estimated using CT images from the Philips Bril-liance 16 CT scanner were 0.29mm (X-axis), 0.51mm (Y-Axis) and 0.90mm (Z-Axis). 3D vector calculations were 0.38 mm for CT and 0.72 mm for MRI. No significant variation is noticed in volume calculations from both image data sets.

Conclusion

This study showed that accuracy and quality of stereotactic CT imaging and MR-imaging are limited by localization devices and their designs in frame based Radiosurgery. This method is simple, direct and economical with good resolution and high reproduci-bility.

Keywords

Stereotactic accuracy, Computed Tomography, Magnetic Resonance Imaging, frame based Radiosurgery

Introduction

The aim of frame based radiosurgery is to deliver a large therapeutic dose of radiation within a three dimensionally defined volume with reference to an extra cranial reference system while minimizing exposure to normal structures outside the target volume. The accuracy of stereotactic radiosurgery is limited by errors involved in each step of procedures. Magnetic Resonance Imaging (MR-imaging) and computed Tomography imaging (CT-imaging) and Angiography imaging are routinely used for stereotactic localization in Radio-surgery procedures.
MR Imaging is used routinely for stereotactic radiosurgery as it is multiplanar and provides excellent contrast resolution between lesion and its surrounding tissues.\(^{(4,11)}\) CT-imaging systems use x-ray beams and utilize differences in attenuation coefficient of body tissues with multiple projection and acquisition methods.

Stereotactic localization requires high spatial accuracy and maximum error more than 1 mm is considered unacceptable \(^{(9)}\) for Radiosurgery. Periodic and routine quality assurance tests performed for CT–Imaging equipment and MR-Imaging equipment may not address specific quality and accuracy needs for stereotactic Radiosurgery. Considering the specific quality and accuracy requirements of these procedure measurement methods with uncertainties more than 1 mm can not be accepted for Radiosurgery. In majority of institutions practicing Radiosurgery, due to limited time for availability of imaging equipment and lack of resources in imaging physics there is a need for designing a simple and reproducible method for addressing these concerns on a routine basis.

**Aim**

The aim of this study was to estimate the stereotactic accuracy of MR-Imaging systems and CT-Imaging systems in frame based Radiosurgery using a simple, direct and reproducible method at high resolution.

**Methods**

A 3D Perspex phantom has been designed which has drilled holes of 1.3 mm diameter and depth varying from 10 mm to 120 mm in eight sectors. Capillary tubes of 1 mm internal diameter filled with copper sulphate solution for MR Imaging and guide wires of 1 mm diameter for CT Imaging are designed. The holes were spaced at 10 mm distances for maximum resolution. The phantom also contained five chambers of known volumes can be inserted with Teflon blocks (for CT imaging) or filled with copper sulphate solution (for MR-Imaging). The plates are connected using a polyethylene rod and tightened with Perspex bolts. Phantom has provision for keeping films, ion chambers, diodes, plates for future investigations. The phantom was attached to the Leksell frame-G (Elekta, Stockholm, Sweden) using a base plate and images were acquired with respective indicator boxes using T1-weighted MP RAGE sequence in Magnetom vision 1.5 Tesla MR-Imaging unit (Siemens, Erlangen, Germany) and helical head acquisitions in Brilliance 16 slice CT unit (Philips, Eindhoven, Netherlands). The parameters used in MR sequences and CT imaging are given in (Tables 1)

The images were exported to the Gamma plan 5.34 (Elekta Instrument AB, Stockholm, Sweden) through hospital network and the stereotactic coordinates of the holes were determined in each slice and are compared with known geometrical values. The volumes calculations performed using delineated images sets of CT and MR separately. The deviations in 3D Vector distances from centre of the Leksell coordinate system (100,100,100) to these five Targets (Fig-2) are calculated and image shift is determined in each image set.

**Table 1:** Imaging Parameters (MR-Imaging and CT-Imaging)
The measurements were repeated 13 times over a period of 9 days and independently verified by two physicists for the concurrency of observation. Each data was acquired for a period of 15-20 minutes for MR imaging and 3-4 minutes for CT-Imaging using routine sequences. The extended time for MR-Imaging was for the optimization of voxel sizes and thus for maximum accuracy.

Results

Mean values of absolute maximum error estimated for MR images from Siemens Magnetom vision MR unit were 0.46 mm (X-Axis), 1.66 mm (Y-Axis) and 2.11 mm (Z-Axis). Mean values for absolute maximum error estimated using CT images from Philips Brilliance 16 CT scanner were 0.29mm (X-axis), 0.51mm (Y-Axis) and 0.90mm (Z-Axis). Errors obtained for CT and MR Imaging units are represented as plots in Fig-4 (A,B,C) and Fig-5 (A,B,C)

Mean value of maximum image shift in CT Imaging was 0.38 mm (X-axis 0.37 mm, Y-axis-0.40mm) and MR-Imaging was 0.72 mm (X-Axis-0.63mm, Y-Axis-0.80 mm). The percentage of maximum error in volume calculations were ‘2’ for CT-Images and ‘2.7’ for MR images. Image shift along X-axis and Y-axis of CT-Imaging unit and MR imaging unit are represented as plots in Fig-6 (A,B) and Fig-7 (A,B).
markers. (6,9,13,17) Many authors have observed large distortions associated with MR-Imaging (2,3,11,16) compared to CT-Imaging. (8) However it is to be noted that the anatomical structure as a base coordinate may introduce serious errors due to anatomical variations occurring from person to person and produces a weaker reproducibility.

In phantom experiments, glass rods (9) or solid acrylic rods (17) of 3-5mm or more (8,9,15) were used.

**Discussion**

Stereotactic accuracy evaluation studies in CT-Imaging and MR-Imaging reported in literature (1-3, 6, 9,13,16,17) were based on either the stereotactic coordinates of anatomical structures of investigated patients (1,2,3,16) or using phantom
As the maximum error acceptable in stereotactic Radiosurgery is less than 1 mm, an uncertainty of 3-5 mm in measurement cannot be accepted. Considering this problem, we have used Targets of 1 mm diameter to reduce the inaccuracies in measurement within 1 mm. As Radiosurgery treatments involve high dose gradients in small volume, the measurement grid size should be small and within measurable ranges. But most of the studies (8,9,16,17) reported in literature used an inter target distance of 15 mm or more for evaluation.

Hence we have used a method with minimal target sizes (1 mm) with high resolution (separated at10 mm) for measurements. These measurements give more confidence in planning and delivering a large dose in Radiosurgery treatments where critical structures lying very close to targets.

**Philips Brilliance 16 CT Unit**

The study showed an increase in error (Fig.4 A) along X-Axis towards middle region (X co-ordinates 100-110) and a decrease thereafter in the stereotactic space. Larger error values are noticed at central region along X-axis. Fig. 6 (A) shows image shift along X-Axis with max-imum error at middle part (Y: 90-110) in Y-axis. Image shift analysis along Y-Axis (Fig.6B) shows minimum error at posterior side and maximum error at middle region (Y-100). This increase in error towards middle region may be due to beam hardening artifacts and adapter support used for stereotactic CT-localization.

Fig.4 (B) shows an increase in error from posterior side to middle region along Y-Axis. Max-imum error is noticed at middle part (Y: 90-110) in Y-axis. Image shift analysis along Y-Axis (Fig.6B) shows minimum error at posterior side and maximum error at middle region (Y-100). This increase in error towards middle region may be due to increase beam hardening artifact towards middle region.

The error analysis along Z-axis (Fig.4 (C)) shows a linear increase in error from superior to inferior direction in stereotactic space. Maximum error in CT-Imaging (1.4 mm) is observed in Z-axis compared to X-axis and Y-axis. This may be due to the presence of stereotactic frame and poor Z-axis efficiency of this CT-unit.

The evaluation shows maximum error in middle part (X-Axis, Y-Axis) and inferior part (Z-Axis) of stereotactic space. But the mean maximum error in all directions was less than max-imum acceptable error for stereotactic imaging. The geometric efficiency of an x-ray beam is the proportion of the total beam that is utilized in the imaging processes. The overall geometric efficiency is divided in to two aspects. The first is the z-axis geometric efficiency, where the proportion of the overall x-ray beam width (dose profile) utilized along the long axis of the phantom is considered. The second aspect, often overlooked, is the detector ar-ray’s geometric efficiency. This defines the proportion of the overall detector area that contains active detector material.

Due to the finite size of the focal spot, an x-ray beam has a reduced intensity at the periphery of the field and this region is referred to as penumbra. If the penumbra were utilized in image production on multislice scanners, the outer detectors would receive a less intense x-ray beam than the inner ones. This would lead to the images from these detectors being narrower and noisier. To avoid these, the collimation on x-ray beam on multislice systems is adjusted such that the penumbra lies beyond the active detectors irradiating them uniformly.

On multi slice scanners, the geometric efficiencies are generally in the range 80-98 % for col-limations of 10 mm and above and 55-75% for collimations of around 5 mm10. For collima-tions around 1-2 mm z-axis geometric efficiencies are as low as 25% on some systems, al-though in dual slice mode they can be much higher.

In the present measurement the collimator size was 1-2 mm which has a geometrical effi-ciency less than 25% which may be another reason for maximum errors along z-axis (mean error- 0.9 mm). The increase in error from posterior to mediol part along y-axis may be due
to metal artifacts or beam hardening. Despite an increased probability of errors at the inferior part of the coordinate system in CT imaging, the mean imaging inaccuracy along Z-Axis is 0.9mm which is acceptable within the accuracy limits.

This study shows dependence on frame and accessories in determining the quality and accuracy of stereotactic localization in each direction. But the mean error observed (0.57mm) with high resolution and minimal uncertainty gives better confidence for planning of high gradient Radiosurgery treatments. The limitation of imaging accuracy along Z axis is a result of technical limitations of equipment and design of the localization systems.

Siemens Magnetom Vision 1.5 Tesla Unit:

As shown in Figure 5(A), MR error increases slowly towards the periphery along X-axis compared to the middle region. The study shows minimum error in X-Direction compared to Y-axis and Z-axis. It also shows an increase in error towards the left side along X-axis compared to right side. This may be due to an increase in magnetic field inhomogeneity towards left side produced by the design of MR-adapter support components. Comparatively low error along X-axis may be due to the fact that phase encoding direction is X-Axis.

Fig.7 (A) shows maximum image phase shift at left side (x-150) in X-axis compared to right side. Minimum image shift (0.4 mm) is observed in the middle region compared to both sides.

The error analysis along Y-axis (Fig.5 (B)) shows larger error in posterior direction (Y:50-80) compared to middle and anterior directions. Maximum error is observed in posterior (Y-50) direction and minimum at the anterior part along Y-axis in stereotactic space.

Fig.7 (B) shows image shift along Y-Axis with maximum image shift at posterior region in good agreement with the error evaluation. This increase in error and image shift from anterior region to posterior region may be due to field inhomogeneity produced by MR-adapter base and table. The decrease in error towards anterior direction in stereotactic space stereotactic space may be due to increase in distance from MR-adapter base and table in stereotactic space.

The error analysis along Z-Axis shows significantly higher values compared to error observed along X-axis and Y-Axis. Fig. 5(C) shows a linear increase in error from superior to inferior direction with maximum error close to the stereotactic frame.

This increase in error close to the stereotactic frame may be due to the design of frame and accessories in addition to technological limitations of imaging equipment. Hence it is clear that magnetic field inhomogeneity introduced by frame based localization systems limits the accuracy of MR-Imaging and thus aims of Radiosurgery.

Apart from tissue dependent chemical shift, susceptibility differences and localization devices, the major hardware-related contribution to geometric distortion in MRI are gradient field non-linearity and static field inhomogeneity. In superconducting MRI systems equipped with sophisticated shimming coils, the geometric distortion due to static field in-homogeneity can be expected to be small when compared with that arising from gradient field nonlinearity.

Mathematical algorithms14 (current circuits) can also be used for correcting some of the above stated problems. Our experience with stereotactic MR-Imaging reveals modification of MR sequences by reducing image voxel sizes reduces MR imaging error within sub pixel level. All these factors were taken to in consideration while doing the measurements. In most of the gamma knife Radiosurgery patients the target is aligned to the centre of the Leksell co-ordinate system (100,100,100) thus centering the MR imaging coil target can further reduce the probability of errors. Further, we started using CT-MR fusion on a routine basis for all radio surgical procedures.
Conclusion

This study showed the stereotactic accuracy of CT (mean error - 0.57 mm) is better than that of MR Imaging (mean error - 1.41 mm). This also showed the accuracy and quality of stereotactic imaging is highly dependent on stereotactic frame and imaging accessories used for localization. The design of MR and CT adapters play an important role in addition to technical limitations in each modality of imaging.

The method is a simple and direct and additionally, it has an excellent resolution and reproducibility for assessing stereotactic accuracy of CT Imaging and MR Imaging in any clinical setup. The phantom with minimal target sizes at high resolution, known volumes with multi imaging compatibility and additional features for Dosimetry is an excellent tool which may be extendable for all modes of radiosurgery procedures. It has better resolution with minimum uncertainties compared with many other methods reported in the literature.

References


4. Erti Adolf, Saringer, Heimberger Karl, Kindl Peter, Quality assurance for the Leksell gamma knife unit; considering magnetic resonance image distortion and delineation fail-ure in the targeting of the internal auditory canal, Med. Phys. 26(2), 166-170, February 1999.


15. Watanbe Yoichi, Lee K Chung, Gerbi J B, Geomatrical accuracy of a 3-tesla magnetic resonance imaging unit in Gamma knife surgery, J Neurosurg(suppl) 105:190-193, 2006
