Introduction

There is still debate concerning the relative advantages and weaknesses of ventriculography-, CT- and MRI-guided stereotaxy for the determination of “invisible” functional neurosurgical targets [1]. CT studies [2–5] have challenged the domination of ventriculography. Some reports [4, 6–9] have shown advantages of MRI, even if the problem of field distortion [4, 10–16] raises doubts. Most of these reports are, however, based on a very limited number of cases and the only more substantial study [7] is restricted to two planes of the three-dimensional stereotactic space. Phantom studies were never performed in the conditions used to assess surgical accuracy. The results are always compared to those obtained with other techniques, making difficult to appreciate the absolute accuracy of the MRI-guided technique.

Abstract

Our goal was to evaluate the accuracy of stereotactic technique using MRI in thalamic functional neurosurgery. A phantom study was designed to estimate errors due to MRI distortion. Stereotactic mechanical accuracy was assessed with the Suetens-Gybel-Vandermeulen (SGV) angiographic localiser. Three-dimensional MRI reconstructions of 86 therapeutic lesions were performed. Their co-ordinates were corrected from adjustments based on peroperative electrophysiological data and compared to those planned. MR image distortion (maximum: 1 mm) and chemical shift of petroleum oil-filled localiser rods (2.2 mm) induced an anterior target displacement of 2.6 mm (at a field strength of 1.5 T, frequency encoding bandwidth of 187.7 kHz, on T1-weighted images). The average absolute error of the stereotactic material was 0.7 mm for anteroposterior (AP), 0.5 mm for mediolateral (ML) and 0.8 mm for dorsoventral (DV) co-ordinates (maximal absolute errors: 1.6 mm, 2.2 mm and 1.7 mm, respectively; mean euclidean error: 1 mm). Three-dimensional MRI reconstructions showed an average absolute error of 0.8 mm, 0.9 mm and 1.9 mm in AP, ML and DV co-ordinates, respectively (maximal absolute errors: 2.4 mm, 2.7 mm and 5.7 mm, respectively; mean euclidean error: 2.3 mm). MRI distortion and chemical-shift errors must be determined by a phantom study and then compensated for. The most likely explanation for an average absolute error of 1.9 mm in the DV plane is displacement of the brain under the pressure of the penetrating electrode. When this displacement is corrected for by microelectrode recordings and stimulation data, MRI offers a high degree of accuracy and reliability for thalamic stereotaxy.

Key words

Magnetic resonance imaging · Neurosurgery stereotactic · Thalamus
A phantom study and a large series of thalamic lesions for the treatment of chronic neurogenic pain and motor disorders allowed us to re-evaluate the accuracy of the MRI-guided technique in all three stereotactic dimensions. The Suetens-Gybel-Vandermeulen (SGV) angiographic localiser was used to determine the mechanical accuracy of the Radionics stereotactic system. These three procedures allowed identification and quantification of the different sources of error.

Materials and methods

The thalamotomy, with its peroperative physiological controls, was approved by the University Hospital ethics committee, and was proposed during the last 7 years to 152 fully-informed patients suffering from chronic therapy-resistant neurological symptoms, mainly in the fields of neurogenic pain and dyskinesias. As operations often comprised exploration of two planned targets and some patients were operated on both sides, a total of 331 thalamic lesions were taken into consideration. However, a number of patients had to be excluded from further detailed analysis. The critical factor was the quality of the postoperative demonstration of the lesion, which could be impaired by movement of the patient during data acquisition or the use of contrast medium at an unfavourable postoperative time (see below). Thus, only 86 lesions in 49 patients were retained for this quantitative analysis.

To determine whether this sample was representative of all patients who had undergone thalamotomy, the distance between the anterior (AC) and the posterior (PC) commissures was taken as a criterion and compared to that of the rest of the patients (103). Figure 1 shows the frequency distribution histograms of AC-PC distances for the two groups. The mean, median, skew and kurtosis for the two histograms indicate a close similarity. This is further supported by a chi-square independence test ($X^2 = 9.49$, df = 7.45; $P < 0.05$) indicating that one cannot reject the null hypothesis that the two populations are the same.

An adaptation to MRI of our CT-guided stereotactic technique [17] was performed with the MRI-related fiducial-based Radionics system coupled to the Cosman-Roberts-Wells (CRW) stereotactic frame. The MRI-compatible head ring contained no ferromagnetic pins, unlike another version [18]. It was fixed to the head under local anaesthesia. With the help of external bone landmarks, care was taken to avoid significant rotation and tilting of the head inside the frame. The plane of the head ring was set between the orbito-frontal and Reid’s plane.

The patient was taken to the 1.3 T MRI unit. A purpose-made fixation plate guaranteed rigid installation of the frame and head inside the MRI head coil. The whole examination was done using T1-weighted images. A rapid series of 5 mm horizontal slices through the dienecephalic area allowed determination of the mid-sagittal plane. Then, a series of 2 or 2.5 mm sagittal slices centered on the midline (TR 450–500 TE 20 ms, 4 excitations, 192 × 256 matrix) was planned. The AC and PC were easily recognised on the mid-sagittal plane, allowing determination of the intercommisural plane and distance (Fig. 2). A series of horizontal slices was then planned parallel to the intercommissural plane, with one slice centred on the desired dorsoventral (DV) co-ordinate of the intended target using a spinecho sequence with TR 450–500 TE 20 ms, 4 or 5 excitations, matrix 256–512 × 256. Slices were 2.5 and later 2 mm thick with a 0.25 and later 0.2 mm interslice gap. The AC and PC co-ordinates determined on the midsagittal plane were transferred to the intercommissural slice for cross-checking (Fig. 2).

The anteroposterior (AP) intended target co-ordinate was then transferred to the appropriate slice. The third, mediolateral (ML), intended target co-ordinate was measured laterally from the border of the third ventricle. To adapt for possible head rotation in the frame, the position of the intended target was always determined on a line orthogonal to the intercommissural line. Head tilting inside the frame rarely exceeded 2°, so that the DV error calculated for this angle and for an average ML co-ordinate of 7 mm amounted to ±0.3 mm.

The fiducial-related technique of the Radionics system was then applied, entering the relative X and Y co-ordinates of the nine localiser rods (Fig. 2) and the intended target in the stereotactic programme to produce the final three stereotactic co-ordinates (AP, DV and ML) of the intended target. The systematic error in AP dimension due to the petroleum oil filling the rods (2.2 mm anteriorly), the AP inaccuracy of the stereotactic material (0.5 mm posteriorly) and the AP MRI distortion (0.4 mm anteriorly) (see below) were corrected by displacing the AP co-ordinate of the intended target of 2.1 mm posteriorly.

The patient was taken back to the operating theatre, and the CRW frame was set with the co-ordinates of the intended target and fixed to the head ring after checking the correct setting of the three co-ordinates on the phantom. The entry point was determined at a precoronal level and, after local anaesthesia, a 2.1-mm-diameter hole was drilled. After coagulation of the dura mater, the stimulation-thermocouple electrode was introduced progressively under impedance control. The uninsulated tip of the electrode measured 4 mm in length and 1.8 mm in diameter (Radionics Gildenberg set). At 10–15 mm before reaching the intended target, the electrode was replaced by a microelectrode for unit recordings. A stimulation session was then performed with the help of a generator. Analysis of all electrophysiological data led then to plan a lesion tailored to each patient, using the Radionics RFG 3C lesion generator, applying a reversible test lesion (45°C for 30 s) followed by one to four definitive lesion units (65–80°C for 60–80 s). Lesion units measured 2–4 mm in diameter and, when more than one, were placed one behind the other to produce a long lesion. This was typically the case for the central lateral nucleus. Radiographs were exposed in the frontal and lateral planes with the electrode at the intended target point or at a known distance from it. Similar radiographs were later obtained with the SGV localiser and without the electrode, allowing off-line application of the SGV software.