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Enhanced co-registration methods to improve intracranial electrode contact localization

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ABSTRACT

Background: Electrode contact locations are important when planning tailored brain surgeries to identify pathological tissue targeted for resection and conversely avoid eloquent tissue. Current methods employ trained experts to use neuroimaging scans that are manually co-registered and localize contacts within ~2 mm. Yet, the state of the art is limited by either the expertise needed for each type of intracranial electrode or the inter-modality co-registration which increases error, reducing accuracy. Patients often have a variety of strips, grids and depths implanted; therefore, it is cumbersome and time-consuming to apply separate localization methods for each type of electrode, requiring expertise across different approaches.

New method: To overcome these limitations, a computational method was developed by separately registering an implant magnetic resonance image (MRI) and implant computed tomography image (CT) to the pre-implant MRI, then calculating an iterative closest point transformation using the contact locations extracted from the signal voids as ground truth.

Results: The implant MRI is robustly co-registered to the pre-implant MRI with a boundary-based registration algorithm. By extracting and utilizing ‘signal voids’ (the metal induced artifacts from the implant MRI) as electrode fiducials, the novel method is an all-in-one approach for all types of intracranial electrodes while eliminating inter-modality co-registration errors.

Comparison with existing methods: The distance between each electrode centroid and the brain’s surface was measured, for the proposed method as well as the state of the art method using two available software packages, SPM 12 and FSL 4.1. The method presented here achieves the smallest distances to the brain’s surface for all strip and grid type electrodes, i.e. contacts designed to rest directly on the brain surface.

Conclusion: We use one of the largest reported sample sizes in localization studies to validate this novel method for localizing different kinds of intracranial electrodes including grids, strips and depth electrodes.

1. Introduction

Intracranial electroencephalography (iEEG) is an important tool for analyzing brain activity because it directly records electrical signals from targeted brain structures. However, there is a need for an improvement in electrode localization to better identify pathological tissue (Smith, 2005; Stefan et al., 1996), spare eloquent cortex (Carmichael et al., 2008; Rodionov et al., 2013), enhance the understanding of neurological disorders on brain function (Misra et al., 2014; Lachaux et al., 2004; Lachaux et al., 2007) and/or improve comparison of data (Carmichael et al., 2010; Hermes et al., 2012; Wang et al., 2012) across populations (Guenot et al., 2001; Dykstra et al., 2012; Klein and Tournelle, 2012).

Methods to improve electrode localization accuracy have been developed (Nelles et al., 2004; Pieters, 2013), yet, they are either limited by the type of intracranial electrode (Morris et al., 2004; Yang et al., 2012) or rely on inter-modality registration which introduces error, reducing accuracy (Morris et al., 2004; Hermes et al., 2010; Taimouri

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The two point sets were automatically fused with a transform, TICP, as a means of transforming one point set onto another. The patient population came from Thomas Jefferson Hospital over the span of three years, with all cases related to intractable epilepsy. Informed consent was obtained and privacy rights were observed in accordance with IRB guidelines. The approach uses three brain images: one before any surgical procedure, the 'pre-implant MRI', and two immediately after the electrodes are implanted, an 'implant MRI' and an 'implant CT'. The pre-implant scan was a structural non-contrast, T1-weighted image acquired using a 3 T scanner (Siemens*), with voxel resolution of at most 1.0 mm$^3$ and field of view (FOV) ≥ 220 mm (Table S1). Immediately after the electrode implantation the implant CT was acquired with voxel resolution of at least 1.0 mm$^3$ and field of view FOV ≥ 256 mm (GE Medical Systems*) (Table S2). The third image, an implant MRI scan was obtained with the same parameters and on the same acquisition machine as the pre-implant MRI (Table S3).

The implant CT slices (DICOM) were converted to standard 3D volume format nifti. The pre-implant MRI volume was processed for surface anatomy using a fully automated protocol, 'recon_all', in FreeSurfer (http://surfer.nmr.mgh.harvard.edu/). The implant MRI volume was normalized (T1 format, white matter intensities scaled to 110 a.u., slice dimensions conformed to 1-mm$^3$ and n = 256 isotropic slices) using a truncated 'recon_all' procedure which also stripped the skull. (Fig. 2: Raw Images (first row)). NOTE – FreeSurfer or other similar programs could be used for these steps. All images were saved in a standard nii format (nifti.nih.gov).

The depth electrodes had inter-electrode distances of 10 mm and contact diameters between 2.0 and 2.5 mm. For grid and strip electrodes, inter-electrode distances were 10 mm and the exposed contact diameter was 4 mm. The thickness of the grid and strip contacts was < 1 mm. Electrodes used in this study were manufactured by either Adtech Medical Instruments (Racine, WI) or Integra Epilepsy (Plainsboro, NJ).

2. Materials and methods

2.1. Data acquisition

Thirty patients (37.5 ± 13.2 years, 18 males and 12 females; 2170 strip, 538 depth, and 364 grid electrodes) with clinical cases requiring the implantation of intracranial electrodes were examined in this retrospective study (Table 1), with implant locations determined by clinical necessity. The patient population came from Thomas Jefferson University over the span of three years, with all cases related to intractable epilepsy. Informed consent was obtained and privacy rights were observed in accordance with IRB guidelines. The approach uses three brain images: one before any surgical procedure, the 'pre-implant MRI', and two immediately after the electrodes are implanted, an 'implant MRI' and an 'implant CT'. The pre-implant scan was a structural non-contrast, T1-weighted image acquired using a 3 T scanner (Siemens*), with voxel resolution of at most 1.0 mm$^3$ and field of view (FOV) ≥ 220 mm (Table S1). Immediately after the electrode implantation the implant CT was acquired with voxel resolution of at least 1.0 mm$^3$ and field of view FOV ≥ 256 mm (GE Medical Systems*) (Table S2). The third image, an implant MRI scan was obtained with the same parameters and on the same acquisition machine as the pre-implant MRI (Table S3).

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2.2. Electrode localization

A boundary-based registration (BBR) method was used to co-register the implant MRI with the pre-implant MRI (Greve and Fischl, 2009). This approach leverages the white/grey matter gradient and the algorithm is robust to signal void artifacts from implanted electrodes, ventricular volume swelling, cerebral edema, and parenchymal shift. The output of this co-registration is a rigid (6 DOF) transformation of the implant MRI volume to pre-implant MRI space, $T_{BBR}$.
Jenkinson et al., 2002) and 'Co-Register' from SPM 12 (Wells et al., 1996; Pluim et al., 2003; Collignon et al., 1995), a mutual information based cost function (Wells et al., 1996; Maes et al., 1997) was implemented with the default settings to co-register the CT to the pre-implant MRI. This approach has been shown to have the best results when compared to correlation ratio (CR) (van Rooijen et al., 2013a).

The output of this co-registration is a rigid (6 DOF) or a non-rigid, initial transformation of the implant CT volume into pre-implant MRI space, $T_{MRI}$. Non-rigid transformation was used to account for the tissue deformation (Studholme et al., 2001) between the pre-implant and implant images. Non-rigid co-registration has been used in other electrode localization studies to register the pre-implant MRI image with the post-implant CT and no additional errors were reported (Hunter et al., 2005; van Rooijen et al., 2013b; Wu et al., 2012). However, the user can also apply a rigid 6 DOF, which will produce fewer free parameters that could alter the co-registration (Table 2).

To identify electrodes in the implant MRI a mask from the pre-implant MRI was used to separate potential electrode contacts from physiological signal voids, the surface anatomy and tissue segmentation obtained from the pre-implant MRI. Using FreeSurfer, the pre-implant MRI was transformed into a binary mask where the zeros represent anatomical voids such as ventricles and empty space and ones represent parenchymal tissue. Due to significant brain shift after implantation, anatomical areas proximal to the ventricles were also masked to zero, including the choroid plexus, cerebellum and brainstem. The mask was applied to the co-registered implant MRI scan to leave behind only those signal voids made by the electrodes (Fig. 3). In this masked implant MRI, the electrode artifacts produced a cloud of 3D points, $A_{MRI}$ that were co-registered in the pre-implant MRI brain space (blue dots, Figs. 2 & 4).

To identify electrodes in the implant CT, the metal artifacts made by the contacts were extracted by binarizing the maximum intensity voxels. The extracted voxels form a cloud of 3D points, $A_{CT}$, that is co-

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**Table 1**

Patient demographics and electrode population. A total of thirty (Greve and Fischl, 2009) patients with surgically implanted subdural electrodes were retrospectively studied. This study examined a variety of cases that included strip, depth and grid type electrodes.

| Patient | Age/sex | STRIP | DEPTH | GRID |
|---------|---------|-------|-------|------|
| TJ018   | 20/M    | 128   | 32    | –    |
| TJ020   | 34/F    | 48    | 36    | 24   |
| TJ021   | 52/F    | 48    | 18    | 24   |
| TJ024   | 44/F    | 36    | 24    | –    |
| TJ030   | 45/M    | 64    | 36    | –    |
| TJ031   | 15/M    | 56    | –     | 24   |
| TJ033   | 53/M    | 56    | 18    | 24   |
| TJ035   | 36/M    | 152   | 36    | –    |
| TJ038   | 48/F    | 128   | 32    | –    |
| TJ040   | 35/M    | 56    | –     | 24   |
| TJ044   | 20/F    | 86    | –     | –    |
| TJ045   | 31/M    | 72    | 16    | –    |
| TJ047   | 50/M    | 128   | –     | –    |
| TJ049   | 44/F    | 72    | 32    | –    |
| TJ050   | 28/M    | 16    | 18    | –    |
| TJ052   | 38/M    | 48    | –     | 36   |
| TJ055   | 56/M    | 40    | 24    | 36   |
| TJ056   | 47/M    | 124   | 32    | –    |
| TJ059   | 27/M    | 112   | –     | –    |
| TJ060   | 20/F    | 40    | 26    | –    |
| TJ061   | 55/M    | 96    | 20    | –    |
| TJ064   | 37/F    | 64    | 6     | –    |
| TJ065   | 57/M    | 48    | 12    | –    |
| TJ071   | 48/M    | 96    | 24    | 24   |
| TJ074   | 48/M    | 180   | –     | –    |
| TJ075   | 30/M    | 80    | 24    | 64   |
| TJ078   | 19/F    | 64    | 48    | –    |
| TJ080   | 24/F    | 104   | 48    | –    |
| TJ081   | 25/F    | 112   | 24    | 64   |
| TJ082   | 57/F    | 72    | 16    | 20   |
| N = 30  | 37.5 ± 13.2 | 2170 | 538 | 364 |

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registered in the pre-implant MRI brain space. When manually identifying electrode centroids from each cloud, there can be inter-observer, and even intra-observer discrepancies (van Rooijen et al., 2013a; Hebb and Miller, 2010). Here, we developed an algorithm to do this rapidly and accurately with minimal human intervention for the implant CT. The suite of analysis code used in this study along with the localization GUI will be available for download under an open GNU license @ https://github.com/Neurorobtics/ACOREL.

The centroid, E, of each cloud of electrode voxels was identified using a brute-force, density-based scanning algorithm (Ester et al., 1996). A human technician is sometimes required to correct instances where the centroid is misclassified due to overlapping electrodes (< 5%). Additionally, this is the most convenient stage to assign recording channel names to each contact centroid (Fig. S1). In the co-registered, post-implant CT volume, the set of centroids of the metal artifacts is another set of points, E_{CT}, in the pre-implant native space (yellow circles, Fig. 4).

The point-sets (A_{CT} and A_{MRI}) are then realigned using an implementation of the iterative-closest point (ICP) algorithm (Besl and McKay, 1992; Kjer and Wilm, 2010). This algorithm seeks to align the 3D points from the implant CT electrode artifacts, A_{CT}, with the 3D points from the implant MRI electrode artifacts, A_{MRI}, using a minimization of least squares of the distances between closest points. In brief, this algorithm computes the rigid transformation, T_{ICP}, that maps A_{CT} onto A_{MRI}, correcting the likely locations of electrodes in the MRI native space using the more accurate location in the CT native space creating a final set of corrected electrode coordinates, E_{MRI}.

2.3. Validation metrics

To validate the approach for strips and grids, the Euclidian distance between the electrode centroids and the nearest location on the pial surface was computed. Since grid and strip electrodes are designed to lie directly on the brain surface, the method should result in a relatively short distance to the surface when compared with other accurate approaches (Crum et al., 2003). The proximity of E_{CT} to the surface is referred to as P_{CT}. For the two most commonly used methods of co-registration used in electrode localization, FSL4.1 and SPM12, the P_{CT} was calculated. The proximities of the electrode location E_{MRI} to the surface is called P_{MRI} (Fig. 4). The population of proximities was compared using a paired-sample, non-parametric Wilcoxon Signed-Rank test (Connover and Iman, 1981). The non-parametric effect sizes are reported as the average of 3 separate “effect sizes” for the non-parametric Wilcoxon signed-rank test. 1. Rank Bi-Serial Correlation (Pallant, 2007; Field, 2016), 2. Simple Difference (Kerby, 2014), 3. Common Language Effect Size (McGraw and Wong, 1992).

To assess accuracy of all approaches for grids, Euclidian distances between the final surface locations and those predicted from the Yang method (Yang et al., 2012) were compared after applying a surface normal projection (Hermes et al., 2010; Taimouri et al., 2013) to the ICP electrode coordinates, E_{MRI}. The Euclidian distance between the electrode localization using FSL 4.1, SPM 12 or ICP and that of Yang et al. (2012) were compared using the median difference.

To understand the significance of the ICP approach and the actual difference it makes in localization, the median distances between all
electrode localization from two different approaches (Methods pairwise: FSL4.1-ICP, SPM12-FSL4.1 and ICP-SPM12) were compared using between group comparisons, separately for grids, strips and depths, for each patient. Data were non-uniformly distributed therefore results are reported as medians ± inter-quartile range. ECT is determined from the standard approach (Hermes et al., 2010) using one of two possible mutual information softwares (FSL 4.1 and SPM 12). The distance (DMeth) between the standard approach centroids, ECT, and the transformed centroids, EMRI, is the Euclidian distance between the two points in three dimensional space, before any surface projection (Fig. 4). We performed a one-sided sign test on the DMeth for each patient, with the medians compared to a typical 2.5 mm electrode radius, i.e. the clinical standard for ECoG accuracy. The ratio of patients who ‘fail’ the sign test by having DMeth > 2.5 mm is used to calculate an alternate common language effect size for clinical relevance.

3. Results

3.1. More proximal localization to surface of grids and strips

For strips and grids, the distance between the electrode centroids and the closest surface vertex should be theoretically less than the thickness of the electrode contact (< 1 mm) because these electrode contacts are designed to rest directly on the brain’s surface. For grid electrode cases (N = 11 patients), the novel method, ICP, yielded electrode centroids that were closer to the surface (PMRI = 4.3 ± 3.2 mm) than either FSL4.1 (PCT = 5.4 ± 3.8 mm, Wilcox Signed-Rank, p = .019 and p = .001, respectively, Fig. 5A), with effects sizes r = 0.5 and r = 0.9. Similarly, for strip electrode cases (N = 30 patients), the ICP method (PMRI = 1.9 ± 0.7 mm) yielded electrode centroids localized closer to the surface than the FSL4.1 method (PCT = 2.4 ± 1.4 mm, p = .002, r = 0.45). However, when comparing the ICP approach to that of SPM12 (PCT = 2.0 ± 1.0 mm), the distances between the electrode location and the brain’s surface were not statistically different (p = .171, r = 0.4). Interestingly, the electrode locations determined using SPM12 were closer to the brain’s surface than that determined for FSL4.1 (Wilcox Signed-Rank, p = .021, r = 0.36 Fig. 5B). This suggests that, if an implant MRI is not available or not warranted, the SPM12 approach is likely to be closer to the surface than the FSL4.1 approach.

3.2. Grid electrode localization is comparable to existing implant MRI based methods

Previously, Yang et al. (2012) developed a method to localize grids that was well validated using photomicrographs of electrode positions on the brain's surface. To assess the accuracy of the novel method, we directly compared the final electrode locations after surface projection to locations predicted by using the method of Yang et al. (2012). The median distance between the locations was 1.9 +/- 2.2 mm, (n = 364 grid electrodes, N = 11 patients) (Fig. 6), i.e. within the diameter of a single electrode (2.5 mm) (Sign test, p = .02).
3.3. Differences in localization methods

The above results suggest that the novel approach is 1) highly accurate and/or 2) equally precise as either FSL4.1 or SPM12. To understand the magnitude of the difference, the distance between electrode localizations were compared across methods (FSL4.1-ICP, SPM12-FSL4.1 and ICP-SPM12) separately for grids, strips (Fig. 7) and depths (Fig. 8). The median difference between the ICP and the SPM12 method did not exceed the diameter of an electrode contact (2.5 mm) regardless of electrode type (Sign test, \( p = .47 \)) (strips = 2.1 ± 1.5, grids = 1.6 ± 1.3 and depth = 1.7 ± 1.0 mm). Conversely, the median difference between the FSL4.1 method and either the ICP or SPM12 method was greater than half the inter-electrode spacing for strips and grids, > 5.0 mm, (Sign test, \( p = .004 \)) FSL4.1 vs. ICP: Strip = 5.6 mm ± 3.9 mm; Grid = 5.8 mm ± 6.2 mm; FSL4.1 vs. SPM12: Strips = 5.9 mm ± 5.4 mm; Grids = 5.2 mm ± 6.1 mm) and as much as 4 mm for Depths (FSL4.1 vs. ICP: 4.1 mm ± 2.4 mm; FSL4.1 vs. SPM12: 3.5 mm ± 2.5 mm). For the depth electrode validation measure of surface proximity, there was no statistical difference between methods, suggesting that not only is the novel method highly accurate for surface electrodes, but it is also accurate for depth electrodes.

3.4. Run times and memory/processing resources for each approach

The run time for the 3D rendered brain surface and cortical reconstruction analysis is estimated using FreeSurfer 5.3, i.e. recon-all; this stage is longest and will typically not be fully completed until the following day, however, FreeSurfer software yields top tier results for surface and cortical reconstructions and many subsequent steps can be automatically run in parallel.

The official FreeSurfer website estimate using “an AMD Opteron 64bit @ 2.5 GHz processor” is 20–40 h, but an informal estimate for our data set, utilizing an Intel(R) Core(TM) i7-3770 CPU @ 3.4 GHz the run-time was on average 10–20 h. All the below steps can be run in parallel with the surface and cortical reconstruction, scaling appropriately for image resolution and processor. For computer memory, each subject’s set of raw dicom scans was about 100 Mb to 200 Mb. Saving all of the image volume files after transformations will result in a folder with total size between 750 Mb to 1500 Mb.

For the Mutual Information step, both SPM12 and FSL4.1 compute the co-registration within 5–15 min and can be done immediately in parallel with the reconstruction. For the post-implant MRI processing there is a skull strip and intensity normalization which take 5–10 min. The boundary-based registration (BBR) step then requires 5–10 min and can be set to automatically start in parallel, but only when the white/grey surface reconstruction step has completed during the recon-all process. After all co-registrations are done, extracting the metal artifacts from both post-implant images (CT and MRI) is instantaneous using intensity thresholding described in the methods. Finally, the ICP registration and ideal transformation calculation takes a mere 5–10 min on average. The optional last step is to project the electrodes onto the pre-implant brain surface via one of several methods, each taking roughly 5–10 min.
4. Discussion

4.1. Discussion of enhanced localization method

Optimized clinical outcomes for neurosurgical patients and related human electrophysiological research both depend on accurate electrode localizations. Here we show a method to improve the localization process by introducing the implant MRI as a source for electrode information relative to soft-tissue anatomy that ensure both surface and depth electrode are well localized with a single approach. The use of the implant MRI provides accurate co-registration results and automatically overcomes inter-modality errors from CT to MRI registrations, as validated by measuring the distance between hypothetical electrode locations and a physiological landmark, i.e. the brain surface from the pre-implant MRI. Moreover, while the metal artifacts from the implant MRI are difficult to distinguish for a human technician (Arnulfo et al., 2015), the artifacts from a standard implant CT clearly distinguishes individual contacts and electrode centroids and the novel approach can be used to automatically assign individual contacts for labeling. By using both CT and MRI implant images the limitations inherent in using a single scan can be avoided because the pre- and post-surgical images are automatically registered with the smallest degree of error. Thus, when implant-MRI is available, this work improves the established method of automated electrode localization by providing a single approach to localize both surface and depth electrodes.

For grid electrodes, after TICP alignment, electrode location resulted in surface proximities that were significantly smaller than uncorrected locations, which relied on mutual information co-registration algorithms from either SPM12 or FSL4.1, (p = .001, Wilcoxon-Sign Rank). A limitation of this approach is that, due to brain deformation around implant sites, the method closest to the surface may not necessarily be the most accurate. However the close proximities for all three methods show them to be in general agreement. By using the void artifacts as landmarks to determine a transformation correction, the errors from inter-modality CT to MRI registration are abrogated. This study's PCT values for surface proximity are close to those in previously reported studies (3.04 mm; Dykstra et al., 2012) (2.3 mm grid only; Laviolette et al., 2011).

When compared to a localization method that also relies on implant MRI for localization of intracranial grid electrodes, e.g. the locations predicted using the method of Yang et al., the ICP method on average was localized within a distance less than a contact radius of a surface electrode (DICP-Yang = 1.9 mm), which is an acceptable range for most tailored surgeries. It is important to note that the semi-automated approach developed by Yang et al. cannot be applied to depth electrodes and must be manually applied to strip electrodes as well as three corner grid electrodes. The differences between the SPM12 method and that of Yang (D_{SPM,Yang} = 2.5 mm) is similar to a previous study that measured the error from CT derived coordinates, (2.3 mm; Laviolette et al., 2011). The ~2.0 mm difference could have been exacerbated by the different surface projection methods. Yang et al., uses an isometric-dilation, i.e. the inter-electrode distances are preserved and the points are radiated out towards the surface from an origin approximately at the brain's center. Whereas, we employed a surface normal projection based on the electrode's relative geometry on the silastic sheet it was implanted on (Taimouri et al., 2013).

The difference between SPM12 and ICP derived centroids for depth electrodes, D_{SPM,ICP} = 1.7 mm, matches the measured difference in CT-MRI derived coordinates from a previous study (1.2 mm at ‘tip’ and 2.3 mm at ‘base’; van Rooijen et al., 2013a, 2013b). This is within an acceptable range to be functionally identical for most excision surgeries. However, when SPM12 or ICP are compared to FSL4.1, the difference is greater than an entire electrode diameter, i.e. D_{SPM,FSL} > 5.0 mm, which can lead to significantly different localizations. It is beyond the scope of this paper to analyze the differences in the FSL4.1 and SPM12 software implementation of mutual information. However, one possibility is that the discrepancy arises during interpolation of the volume after the co-registration transform is applied. Alternatively, it might be due to the fact that FSL4.1 uses different iterations to maximize the mutual information cost-function. There is one formal report (van Rooijen et al., 2013b) that mutual information was the best cost function when compared to correlation ratio and least squares, however this is the first comparison between different software implementations of mutual information algorithms.

4.2. Limitations

In any image co-registration paradigm, the registration error can be corrected through manual adjustments by an experienced technician; however, this process is prohibitively time consuming and studies will still suffer from inter-user variability. This is why the novel method strives to maintain as much automation as possible, thereby obviating the need for a trained technician. However, the enhanced process is not completely automated. Semi-manual identification of each electrode centroid to a specific recording channel is still necessary (Fig. S1). Briefly, after the post-implant CT and pre-implant-MRI are co-registered, then a trained expert must use a combination of visualization software and hand-taken notes from the operating room to meticulously and precisely assign a centroid to a single voxel from the co-registered CT scan metal artifacts along with its corresponding channel. The process takes anywhere from 2 to 4 h depending on the total number of electrodes implanted in the patient, which can typically range from 50 to 150.

From Table 3, we see that the enhanced method while employing two extra steps does not require significant more run-time and the critical bottle neck is actually the above centroid-channel correspondence step which can itself be fully-manual or semi-automated. There are two extra steps in the enhanced method that add 10–20 min, however the centroid-channel correspondence step is semi-automated in the enhanced methods, so this indeed saves 1–2 h.

The bulk of signal voids are accurately located in the implant MRI, however, the edge of the contacts are distorted by the artifact's

| Table 3 | Localization approach steps and run times in parallel layout. Run times are shown for one typical subject and 1.0 mm³ resolution MRI scans and 0.5 mm³ CT scan; the files will occupy roughly 1.0 Gb of hard disk space. Steps are ordered from first (top) to last (bottom). |
|---------------------------------|---------------------------------|---------------------------------|
| Surface and cortical reconstruction (FreeSurfer5.3) | Standard (SPM12 or FSL4.1) | Enhanced (ICP) |
| Recon-all (10–20 h total) | White/Grey surface complete (8–12 h) | Mutual Information registration (5–10 min) | Intensity normalization + skull strip (5–10 min) |
| | Spherical surface registration to atlas (2–4 h) (optional) | Extract artifacts via intensity; maximum | Boundary-based registration (5–10 min) |
| (optional) Cortical Parcellations (5–10 min) | Manual centroid-channel-correspondence (2.0–4.0 h) (optional) | Surface Projection (5–10 min) | Extract artifacts via intensity; less than grey matter |
| Total: 10–20 h | Total: 2.25–4.33 h | Total: 1.33–1.66 h | ICP registration (5–10 min) |
| | | | Semi-automated centroid-channel-correspondence (0.5–1.0 h) (optional) | Surface Projection (5–10 min) |
penumbra, enlarging the electrode's shadow by up to three times its size (Hoffmann et al., 2008). Recently, it has been shown that with proper MRI sequence settings the distortion can be reduced to less than double the actual contact size (Sarkar et al., 2014). Therefore, the ICP approach, applied with proper MRI sequencing should still be highly accurate for surface and depth electrodes.

5. Conclusion

For intracranial EEG analysis, it is beneficial to have the most accurate electrode localizations possible with little to no manual intervention necessary. The proposed localization method is automatic and accurate because it uses signals from the implant MRI as the ground truth for electrode positions relative to soft-tissue anatomy. Importantly, the two other scans, pre-implant MRI and implant CT, as well as multiple image processing software packages are still required to localize electrodes to corresponding brain morphology. The implant MRI scan has limitations such as artifact size and discriminability as well as safety considerations using magnetic resonance equipment with metal electrodes. However, using the complete set of images in the enhanced manner shown here, the individual limitations from each scan are effectively circumvented and a single, automated approach can be utilized with comparable accuracy to existing methods.

Disclosure

The authors have no conflict of interest to disclose and there were no conflicts of interest in this study.

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Appendix A. Supplementary data 1

Supplementary data to this article can be found online at https://doi.org/10.1016/j.nicl.2018.07.026.

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