# Fusion and display of 3D SPECT and MR images registered by a surface fitting method

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### Abstract

Since 3D medical images such as SPECT and MRI are taken under different patient positioning and imaging parameters, interpretation of them, as reconstructed originally, does not provide an easy and accurate understanding of similarities and differences between them. The problem becomes more crucial where a clinician would like to map accurately region of interest from one study to the other, by which some surgical or therapeutical planning may be based.

The research presented here is an investigation into the problems of the registration and display of brain images obtained by different imaging modalities. Following the introduction of an efficient method some clinical useful application of the registration and superimposition were also defined.

The various widely used registration algorithms were first studied and their advantages and disadvantages of each method were evaluated. In this approach, an edge-based algorithm (called surface fitting), which are based on a least-square-distance matching, were suggested for registering of brain images. This algorithm minimizes the sum of square-distances between the two surfaces obtained from two modalities. The minimization is performed to find a set of six geometrical transformation parameters (3 shifts and 3 rotations) which indicate how one surface should be transformed in order to match with the other surface.

**Key words**: Medical image registration, 3D data Matching, Surface Fitting, Super imposition and display, SPECT imaging.

### Introduction

The original 2-D images (from different modalities) are collected under different imaging parameters and system characteristics. In order to display the relevant information between the two studies in a clinically useful form (such as aligned planes), Superimposition and even side-by-side display of the images should be done in a similar matrix size, identical orientation, zooming, and scaling, and nearly similar gray level range. To achieve these objectives, the geometrical differences (e.g.; shift and rotation) between two set of images should be found, then, one set can be spatially transformed to be matched with the other

The following limitations can be considered as the key problems to the registration of 3D medical images:

1) The routine medical imaging (MRI, SPECT, CT) provides adjacent 2-D gray-scale slices of a few millimeters thickness (e.g. 6 to 10). In some cases (e.g. MRI), a gap is introduced between two adjacent slices. In order to access the entire feature in one study corresponding to the other, 3-D information is required. Grey level interpolation is the common key to obtained the gray level of the structures within the gap region. For thick slices, which contain fine features, some uncertainty might be presented in the

definition of the features in the interpolated region between the two slices. The problem can also be overcome by using a shape-based interpolation (r ef) by which a set of 2D binary segmented images are interpolated for a connected 3D external surface.

2) The original images formed by medical imaging systems are influenced by system's physical characteristics such as noise level, contrast and brightness. The images should be processed to improve their visual appearance and thus enhance their clinical interpretation. In this respect, different methods of image enhancement (e.g. digital image filtering) might be used. However, there is no unique effort to improve the quality of an image with regard to different observers. Moreover, different applications might imply different types of image enhancement.

3) As mentioned above, a set of transformation parameters needs to be obtained as a result of a registration process. These parameters are used as input to a desired image reformation algorithm and display system. This coordinate transformation is given in respect to either a reference coordinate system recognized by either modalities, or the coordinate system of one of the modalities. The aim of the superimposition process is to display both images of an object in one common coordinate system. In order to calculate the position of an image pixel being originally on any coordinate system and to display it onto a display system, the pixel point must be transformed from its own coordinate system [xw, yw, zw] into the eye or screen coordinate system [xs, ys, zs].

[xs, ys, zs]=[xw, yw, zw][TR] Equ. 1

The transformation matrix [TR] may be built up from several rotations and translations (shifts), determined by the registration process.

In terms of display types, the most straightforward method is to display the 2-D images from different modalities side-by-side on the screen. The original gray-scale image of one modality (destination image) can be displayed together with the resampled slice through the data set of the other study (see e.g. figures 9.6, 9.7). The decision of the choice of destination image is influenced by the clinical use and the type of data sets.

## Materials and Methods

A registration technique based on surface fitting was used to register MR with MR, SPECT (HMPAO) with MR, and CT with MR brain images of a number of patients. Skin surfaces (scalp) were used when data set involve MR and CT images, whereas, for registration of SPECT brain images with MRI, the surface of brain was used. The characteristics of the imaging systems and the routine imaging parameters used in the current project are as follow.

MR scans were performed with a 1.5 Tesla system operated. 2D slices was obtained by a 3D FLASH (Fast Low Angle Shot) method. Different imaging sequences and parameters were used, creating T2-weighted, T1-weighted and STIR (Short Tau Inversion Recovery) images. A pixel size of 0.70-1 mm (300 mm field of view, matrix size of 256*256, slice thickness of 5-6 mm, and a gap of 2-2.5 mm (when STIR sequence is used) was set in most studies. Acquisition was typically done with TR=3.0 sec and TE=90 msec for T2, and TR=600 msec and TE=15 msec for T1-weighted images. Transaxial slices were usually obtained starting few millimeters (about 10 mm) from the top of the head, down to the base of skull. For the detection of skin surfaces, a short echo time was applied. STIR sequence was used to create 2D fat-suppressed MR images, which makes the detection of the brain surfaces easier.

SPECT images were created using 99mTc (Technetium) labeled radiopharmaceuticals. The actual resolution of the system was about 8 mm at FWHM (at 10 cm in depth). The data were acquired after injection of HMPAO (HexaMethylPropylene Amine Oxime) and transverse scans were reconstructed with a slice thickness of a size similar to pixel size. A matrix size of 64*64 gives a pixel size of about 6 mm at a zoom factor of 1, which was used to obtain 2D slices. External surfaces from SPECT can be defined as the outer envelope of the emission distribution detected by setting of a threshold value. However, decisions on threshold value and the type of threshold selection technique are very important factors in the accuracy of the detected brain surface.

Various methods for registration developed by other authors were reviewed and compared, first. Many of these are based on registering external reference markers, and are cumbersome and cause significant problems to both patients and operators. Internal markers such as anatomical points have also been used, but these may be very difficult to identify. Alternatively, methods based on external surface (surface fitting) of scull or brain have been developed which eliminate some of the problems associated with the other methods.

Among different registration techniques, surface fitting was used to register two sets of brain data obtained under different imaging modalities (MR and SPECT). The fundamental grounds used to select this registration method were: its ease, applicability in routine imaging practice, and power of the algorithm to achieve the result regardless of the internal structure of the object. The surface-fitting algorithm was first implemented by Pelizzari and Chen 1987 on brain medical images. The basic concepts of the method remain the same in this project. However, an attempt was initially made to overcome all the obstacles believed to exist in this method. Some new approaches were also suggested mainly with respect to minimization algorithm to facilitate the technique for use in the routine clinical applications.

In this approach, the surface of an anatomical structure is used to describe the patient-specific geometrical information inside the coordinate system of each imaging modality. These models of surfaces can be transformed geometrically in a spatial Euclidean (Cartesian) coordinate system in order to be brought together into a 'fit' position. Since anatomical surfaces (e.g. brain or head) are not symmetric (such as sphere or cylinder), a unique transformation is expected to match the two surfaces. It is assumed that these surfaces and the images from which they are derived can be scaled to the same size based on the prior knowledge of data acquisition and imaging parameters, and thus scaling is not a part of the fitting procedure. Correction of both data sets for pixel size prevents any geometric inconsistency, which might lead to misregistration.

### Algorithm

The original 2D or 3D gray level image data sets are processed and then geometrically scaled in a preprocessing stage. The external surfaces of both registering objects are generated by application of a surface detection algorithm on serial slices of the pre-processed data set. Then, the binary surfaces (being the set of voxels lying on the surface as defined by the surface detection algorithm in 3D space) are used as primitives to be registered.

The surface to be fitted is referred to as objective surface (Po), and the surface which is transformed at each stage of the fitting process is called transforming surface (Pt). For the objective surface, a 3D binary surface (solid model) is used, but for the transforming surface, in order to save computation effort, only a set of points on the surface are selected for fitting. The distances between these sample points and the surface are then measured as the mismatch value between the two surfaces (see figure 1). A Least Square Distance (LSD) function is defined to evaluate this distance and to minimize the misregistration between them.

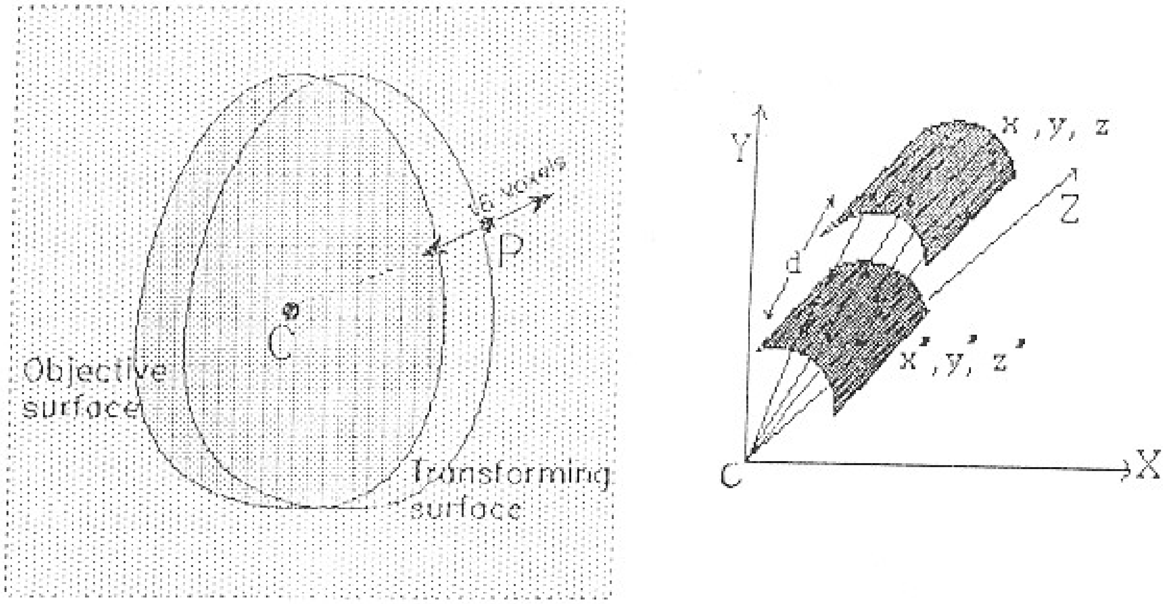


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**Figure. 1-** (a) Schematic diagram showing two fitting surface (ie; objective surface of MRI data and transforming surface of SPECT data set), and its cross section (b) (contour of 2D slices) showing the distance (LSD) between corresponding points of the two surfaces, obtained by ray tracing from sample points on surface to centroid (C) of data set.

Figure 2 shows a block diagram of the surface-fitting algorithm as used in the current work. As shown in this diagram, the algorithm consists of several steps: surface detection, centroid and manual registration, point sampling, point correspondency (using an intersection process), distance measurement, visual inspection, minimization and finally evaluation of the mismatch between the two surfaces. In the first step of registration, the centroid of both binary surface images are calculated and transformed to the origin of coordinate system of the objective surface (centroid registration). Alignment of two centroids brings the two surfaces into a close position and make the process easier to measure the distance between the two data sets. Manual registration is also facilitates automatic registration process. The knowledge of system geometrical parameter information,

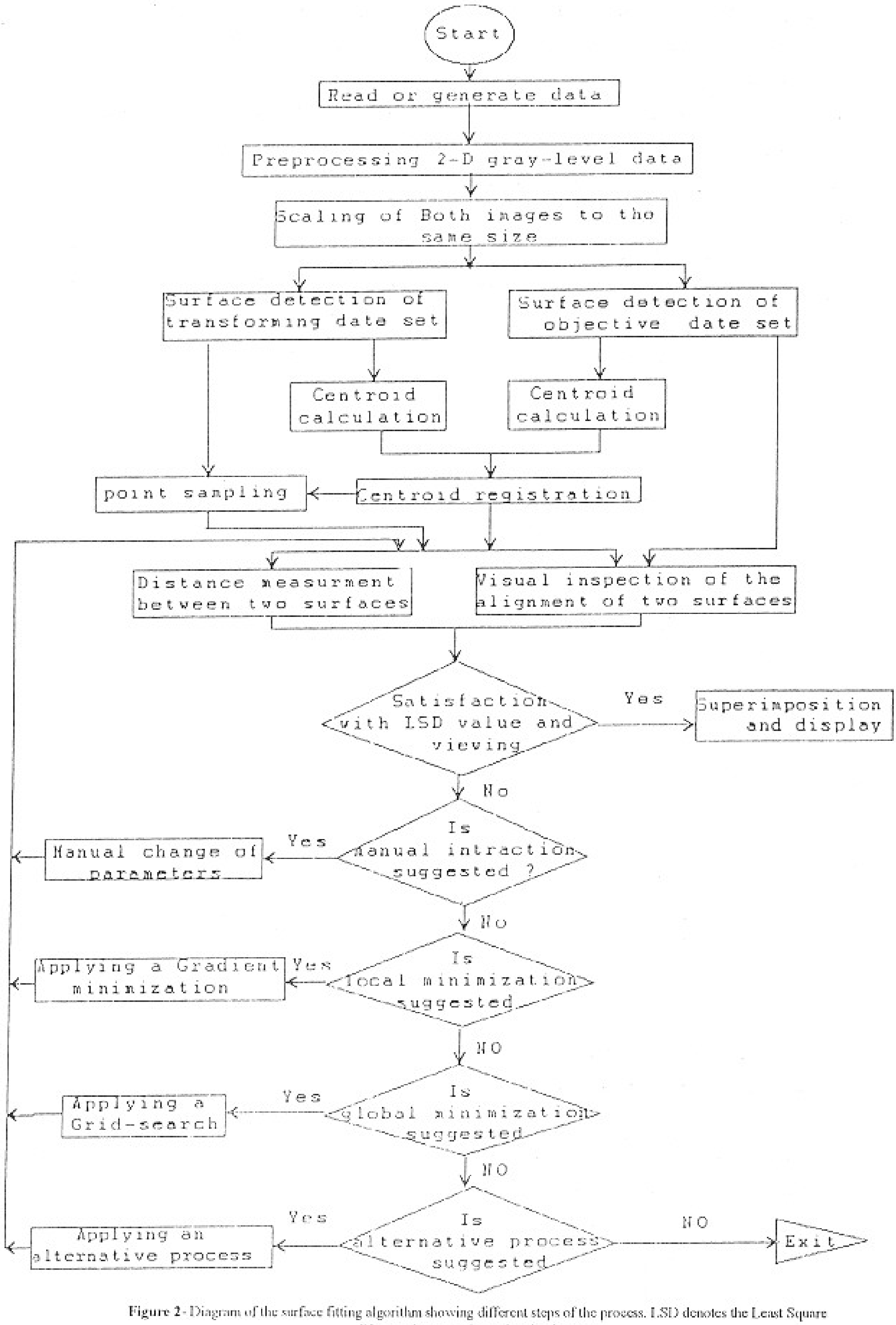


**Figure 2**- Diagram of the surface fitting algorithm showing different steps of the process. LSD denotes the Least Square Distance between the registering images.

and visual judgement of overlapping surface images on viewing screen are essential for proper manual positioning and registration.

To perform the distance measurement, for each sample point of the transforming surface, a corresponding point should be determined on the objective surface. The corresponding point is assumed to be at the intersection point of a ray originated at the centroid, passing through the sample point of the transforming surface, and intersecting with the objective surface. As mentioned above, the fit is then evaluated by a L2 norm least squares distance function (LSD), and minimized in respect of the best transformation parameters. Accordingly, the mismatch value (LSD) is the sum of the squares of the distances between each individual pair of corresponding points on the two surfaces. Individual distance errors can be accumulated as sample points are examined and this type of error is known as cumulative distance error (CDE).

In order to minimize the distance function in respect to changing transformation parameters (3 shifts and 3 rotation in x, y, and z direction as function variables) two types of local minimization algorithms: the direction set method of Powell, and conjugate gradient method of Fletcher-Reeves (e.g. FRPRMN) was compared. As shown in this project, the process may not converge to a global minimum, due to an early detection of a local minimum. This response depends highly on the relative location of the two registering surfaces from which the minimization starts. This is the main disadvantage of the local minimization process since it is user-dependent and relies on the manual interaction of operator.

In the current work a number of improvements was performed to the general surface-fitting algorithm. One of the key improvements is due to the use of a global minimization algorithm (checking any possible transformation parameters for the best fit) which increase the accuracy of the registration. In this approach, a strategy to achieve the registration in a reasonable duration of time owing to two novel approaches, sequential and multi-grid registrations defined in literature (Oghabian M A and Todd-Pokropek A).

The accuracy and cost of applying the surface-fitting algorithm were also evaluated. The accuracy can be defined as the ability to fit the surface images with an acceptable minimum mismatch value. Measuring accuracy is easy on phantom images, but the results are not exactly relevant to actual clinical images. In the current work the data from multi modality imaging of two different types of phantom, the Hoffman brain phantom, and the Jaszczak phantom were used to assess 2-D and 3-D registration, respectively. The mean LSD values and standard deviation after fitting process was 1.44mm and 0.42mm, respectively, for MR-MR registration, and 3.17mm and 1.12mm for MR-SPECT registration. The clinical data sets used for this evaluation are based on two known arbitrary misregistered patient data sets. To simulate well known SPECT images, the original MR images were degraded by a Gaussian filter to nominal resolution of SPECT system and Gaussian noise was added to them. It was shown in this experiment, that the accuracy obtained by MR-to-MR registration was of the order of 1 voxel (corresponding to a transformation error of 1 voxel shift or 1 degree rotation). This accuracy was degraded to about 2.5 voxels for MR-to-SPECT registration (which corresponds to an error of about 2-3 voxels in shift, or 2-3 degrees in rotation). The timing response of Powell method was better (which is about 300 seconds) than it is for the FRPRMN's (about 360 seconds as averaged).

The accuracy of real clinical data sets was also obtained by setting four external markers on the head in both studies (i.e. MR and

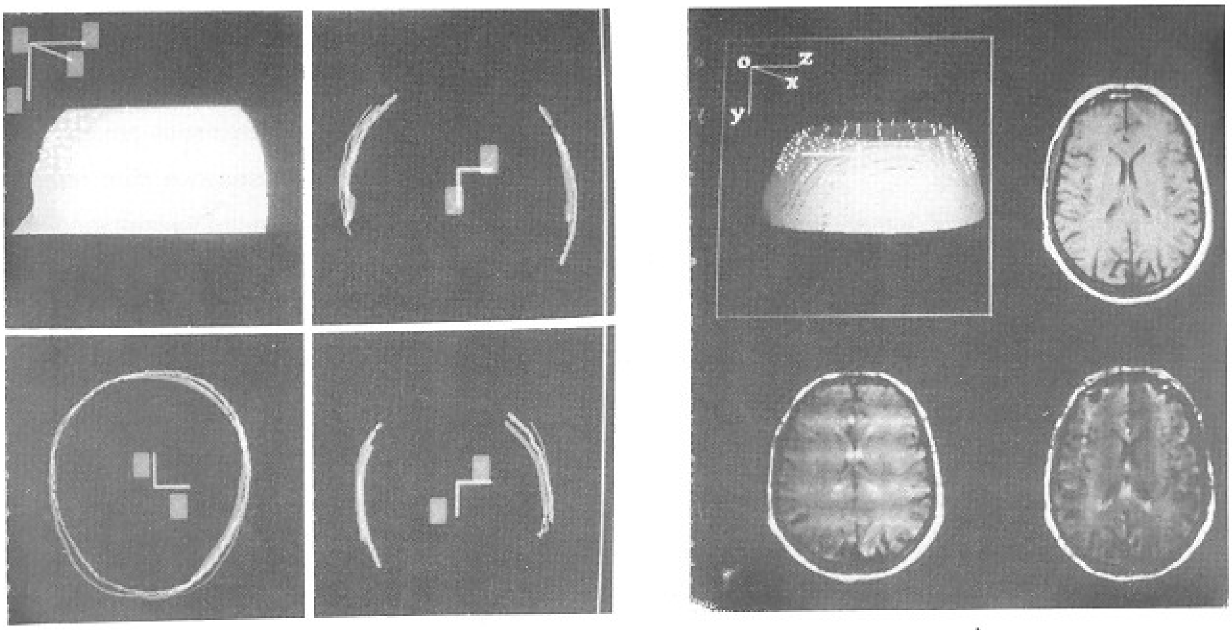


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**Figure 3**- (a) Shaded surface display and cross sectional contours in three orthogonal planes for MR-MR registration during surface fitting process, (b) Registered 2D images displayed side-by-side. the sample points from transforming surface are overlaid on 3D surface of the other registering study.(on top left corner).

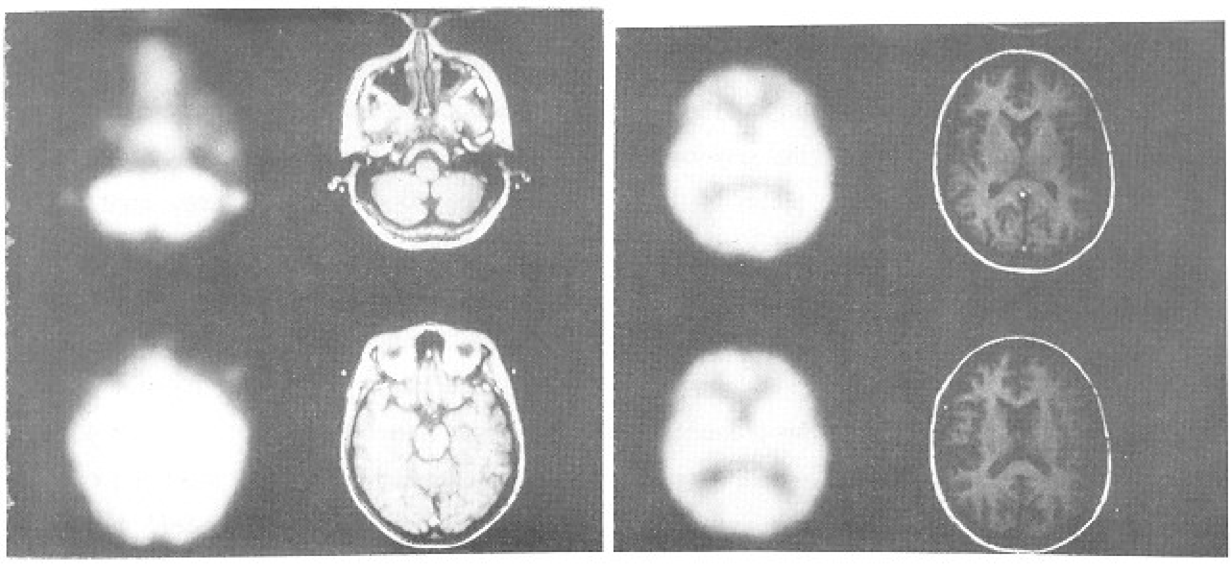

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**Figure 4**- (a, b) MR-SPECT registered images displayed side-by-side. Four landmarks which are visible on 'a' were used to verify surface fitting algorithm.

SPECT) and the results from 10 such data sets were assessed.

This multiple measurement of the registration value shows whether the distance values obtained by the surface fitting process are due to misregistration (displacement) error, or other type of errors.

The accuracy obtained by registration of MR and SPECT data using multi-resolution surface fitting algorithm was 3.47mm with SD of 1.17mm.

The problem of image superimposition (overlaying and fusing the correlated images) and display were also assessed. The 3D surfaces displayed in the current work were based on voxel-based representation methods to shade surfaces and display them realistically during superimposition process. To improve visualization, the volume rendering process was also used for displaying the skin surface and the surface of some internal structures of the head (e.g. brain, tumor and ventricles).

Four methods for fusing and displaying the correlated (registered) images were employed. The correlated 2D slices from one study were created by resampling their original gray scale data oriented along the scan planes of the other study. The side-by-side display of these correlated images was then employed to provide corresponding clinical information between the two studies. The contours of regions of interest (e.g. ventricles or tumors) were superimposed on 2-D gray-scale images of the other study. A linked cursor was used, moved manually by a mouse digitizer on the screen, showing the corresponding structures between the two images. A color-coded display was also suggested in which different color ranges (e.g. look-up tables) are assigned to the structures (e.g. tissues) of each image.

**Results and Discussion**

Although the registration process was the main objective of this work and showed to need much effort for a proper fit, without a powerful display strategy it is not possible to achieve valuable diagnostic information. In order to produce a 2D-registered slice for display purpose, the reslicing was performed by a trilinear gray-scale interpolation process. The coordinates of the destination slice are transformed by the resulted registration parameters in order to obtain the coordinates of the resliced image. Due to the differences in gray level range and size of the registered images, contrast enhancement and linear image scaling should be further employed for a good interpretation of the images on a viewing screen. Image filtering (e.g. median filtering) for noise reduction and color coding needs to be done as part of image enhancement to improve diagnostic ability.

The 3D shaded display of external surfaces (head or brain) as shown in figure 3, helps for understanding 3D spatial position of the surfaces during surface fitting process.

3D shaded surfaces can also be used to show the 3D relation of various displayed objects (e.g. tumor and ventricles) in superimposition (fusion) process.

Typical side-by-side display of the registered images is shown in figure 4. The regions of interest (ROI; e.g. tumor) were also selected manually from MRI study and their contours, as demonstrated in figures 5, 6, were superimposed on 2D gray-scale SPECT images. As shown in these figures, the contours of some specific structures such as brain ventricles in MRI were superimposed on SPECT slices. A linked cursor as shown in the bottom lower half of figures 5 and 6, can be moved, manually, on screen by a mouse digitizer to show the corresponding structures between the two images. Alternatively, the ROI from one modality can be mapped on the context of the other image and displayed in different color coding. The lines passing through shaded surfaces (see top left images of figures 5 and 6) show the position of destination slice on display screen.

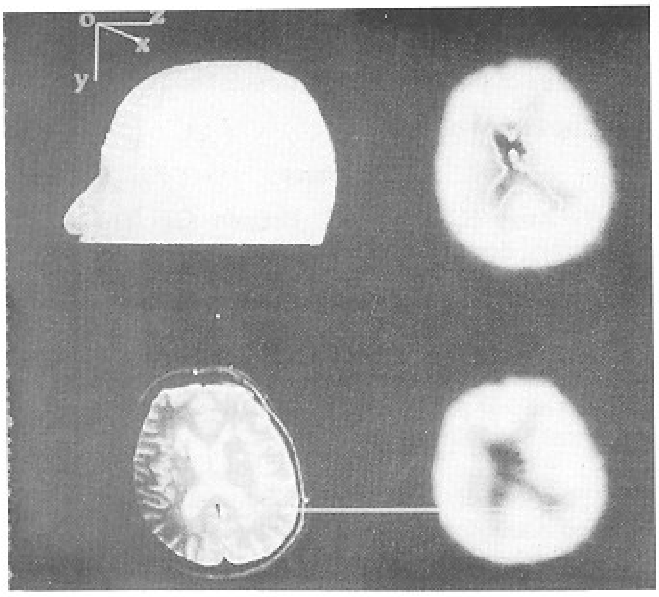


**Figure 5-** MR-SPECT correlated images showing MR contour of ventricles superimposed on resliced SPECT image. A linked cursor is displayed on the bottom half of the picture.

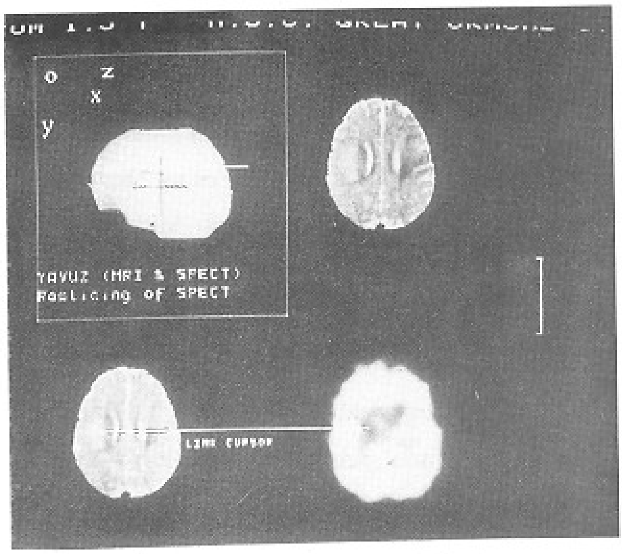


**Figure 6-** MR-SPECT correlated images of brain of a 7-month old baby after registration process. Side-by-side display of original MR slice with resliced SPECT image and linked cursor are also used.

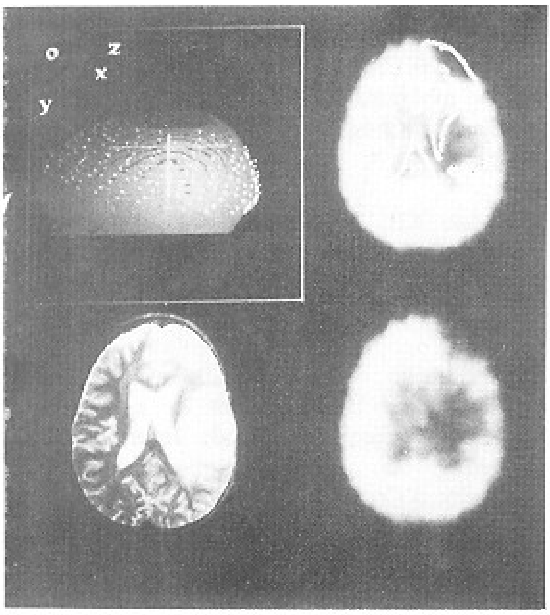


**Figure 7-** MR-SPECT correlated slices and 3D registerd surfaces displayed after registration process. Tumor and ventricle contours from original MR slice are superimposed on resliced SPECT data

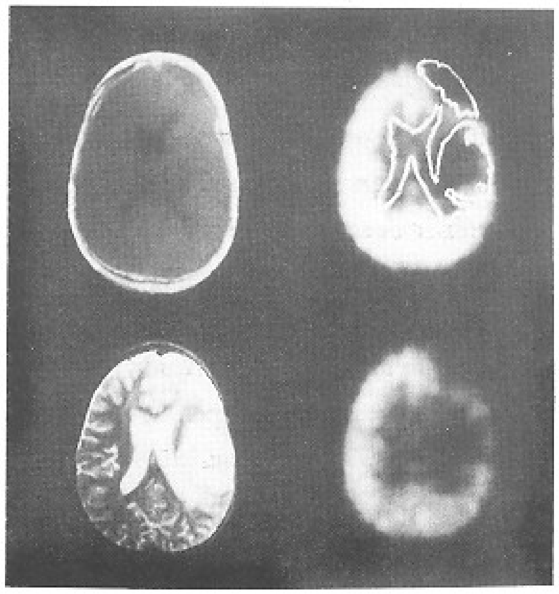

**Figure 8-** MR-SPECT-CT correlated slices displayed side-by-side after registration process.

The registration process was performed on the images of twenty different patients. Most of the patients were children of age less than sixteen years old. The cases were selected from the patients whose images showed an abnormality in one or both modalities. All scans of each patient were done within a maximum period of two months.

The MR-SPECT correlation shown in figures 4 was obtained using the proposed fitting process on axial MR slices and HMPAO SPECT images. Four external landmarks were also used (see figure 4a) for landmarks registration to verify the surface fitting algorithm. Both experimental results and visual judgement of displayed correlated images confirmed reliability of the process. Figures 6 show the MR-SPECT correlated image of a 7-month old female suffering from seizures. Axial double echo STIR sequences were performed to obtain MR images, which were correlated with HMPAO SPECT data. MR data shows high signal in white matter alongside the body of right lateral ventricle. There is also a diffuse area of increased signal in the right temporo-parietal lobe involving gray matter. As shown in these figures, the low uptake area in SPECT images corresponds to the abnormal area of MRI.

The images shown in figures 7 and 8 were obtained from a 4-year male patient with an extensive abnormality of left frontal and left fronto-parietal regions. The axial T2-weighed MR images show extensive abnormalities in white and gray matter of these regions correlated to both CT and SPECT images. Involvement of posterior limb of the left internal capsule, and a lesion in the right striatum and parietal occipital region is demonstrated in this image. In one part of these images, the contour of some lesions which are well visualized in MR images has been placed in the context of SPECT slices acquired under uptake of HMPAO. As shown in the correlated SPECT image, multiple abnormal signal areas are observed, consistent with ischaemic lesions. This suggests that the nature of these multiple infarcts is in vascular distribution of the brain.

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