Image processing and registration to quantify and rectify MRI distortion

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Abstract

Brain diseases affect a sizable of people worldwide. The World Health Organization says that almost 0.004% of the world population (about 256 213 people) suffer from them and 1.8% of cancers diagnosed are brain tumours. Brain disease is difficult to either diagnose or treat. Its diagnosis relies on medical imaging techniques such as magnetic resonance imaging (MRI) and computed tomography scanning(CT), that allow us to observe inside the brain and guide the procedure of surgery. Typically, neurosurgeons use MRI images due to their high spatial resolution and the possibility of detecting changes in brain morphology to plan the trajectory of the operation. However, MRI produces an intrinsic distortion; in brain surgery, even small distortions can lead to fatal mistakes. Although others factors also change the brain's anatomy, the focus in this thesis is on quantifying the MRI distortions compared with CT images which do not have any significant distortion compared with the real object. For that, a grid phantom was scanned on two different MRI scanners and on one CT scanner to assess variations in the geometric distortion in MRI images.

This thesis deals with image registration methods based on features and on images intensities in order to quantify and rectify the MRI's geometric distortion. To carry out the registration based on the features, imaging processing methods are used to extract the features from CT and MRI images and create different point clouds. The registration methods used in this thesis are the Rigid, Affine and Non rigid. The Rigid registration method quantifies the real MRI distortion once it only rotates and translates the objects. Its average distortion remain between 1 and 1.6 mm. Although the Affine registration deforms the objects, due to the nature of the distortion, its results do not diverge from the rigid registration results. The best result is obtained with the non Rigid registration which deforms the object with many degrees of freedom. This method almost completely rectifies the MRI distortion to 0.0086 mm.

Key Words: Image, magnetic resonance(MR), computed tomography,image processing, point cloud, registration.

1. Introduction

In the Post-modern Age, there has been an extraordinary scientific and, more specifically, technological explosion, led by informatics and robotics. It is in this context that the acquisition, processing and analysis of digital imaging overcomes the limitations of human vision regarding the visible band of the electromagnetic spectrum, making it possible-through the use of ultrasound, X-ray, magnetic resonance and computerized tomography, among other acquisition methods - to capture images that allow us to conceptualize a universe that it would once have been impossible to unveil. For instance, there is the detection of different types of materials in the ground, searching for drugs, biometric recognition and, the most impressive adventure yet experienced by man, the voyages inside a living human body, which enable the diagnosis of diseases and surgical and therapeutic procedures, improving the quality of life of thousands of people, and even reduce the number of fatalities associated with complex surgeries.

The aim of this study is to present the methods used for the acquisition, processing and fusion of medical images in order to calculate possible distortions so that the images may provide the doctors that use them with quantitative, objective, reliable and reproducible information and help with medical diagnoses and the accurate locating of lesions.

2. Background

The use of medical imaging has increased over the years and seeks to reveal internal structures hidden by the skin and bones, as well as, to diagnose

and treat diseases. Moreover, it can also be used to establish a database of a normal anatomy and physiology to identify abnormality. There are different tissues and structures inside the human body, so different specific methods to acquire these tissues or structures are needed, each of them with their own advantages and disadvantages.
Despite their vastly different frequencies and principles of operation, there are common steps in signal processing of signals received by these imaging detectors, such as signal amplification, filtering, multiplexing, and signal converted into electrical current. The medical imaging techniques used in this work are: magnetic resonance imaging (MRI) and computed tomography (CT) which are used to support the neuronavigation procedure.

2.1. Magnetic Ressonance Imaging(MRI)

Magnetic resonance imaging (MRI), also known as nuclear magnetic resonance imaging, is a medical diagnostic technique for creating detailed 3D images of the human body using the principle of nuclear magnetic resonance. The mainly MRI components are primary magnet, which produces a very powerful uniform magnetic field, gradient magnets, which are much lower in strength, RF coils and a very powerful computer system, which translate the signals transmitted by the coils[12]. The primary magnet is responsible for align the hydrogen molecules inside the human body while the gradient magnets produces a variable magnetic field, in the radiofrequency range, that excites the atoms, changes their motion and causes a phenomenon called Nuclear Magnetic Resonance[12, 9]. The motion change cause an electric charge that is detected by the equipment and produce the image. The images are stored in a standardized format file called DICOM.

2.2. Computed Tomography (CT)

Computed tomography (CT) is a method of acquiring and reconstructing the 3D image of a thin cross section on the basis of measurements of X-ray attenuation. In comparison with conventional radiographs, CT images are free of superimposing tissues and are capable of much higher contrast due to elimination of scatter. Despite its internal complexity, a CT scanner is basically a set of rotating X-ray tubes and radiation detectors[12, 6]. The basic principle behind CT is to take a large number of X-ray images at multiple angles and, based on that information, calculate the 3D image. To better understand how a CT scanner works it is necessary to understand how X-ray works and the concepts are basically the same. X-rays have a very short wavelength that are able to pass through most tissues inside the human body and as much the beam penetrates into the matter as much the intensity of an X-ray beam decreases, this is called attenuation. The attenuation depends on the tissue[7]. For example, the soft tissues have a low attenuation coefficient, whereas the bone has a particularly high attenuation coefficient. CT scanners makes a lot of attenuation thorough the plane of the thickness cross section of the body and use this data to reconstruct a digital image of the cross section. The images are also stored in a standardized format file called DICOM.

2.3. Neuronavigation

Image guided surgery has become an essential tool in brain tumour surgery[11]. The head, the skull, the skin do not show us where the tumour is. In the old days, before image guidance, the doctors would have to make a very large opening in surgery in order to find the tumour. Today, the doctor can take this information through the MRI and CT images, which can load into a computer that provides us 3D images. This system is very similar to the GPS system for cars, which allows us to navigate around the town and find a spot. The GPS image guided surgery allows the doctors navigate around the brain and find the tumour[11]. As with GPS, there are some very important components: camera, pointer and computer.

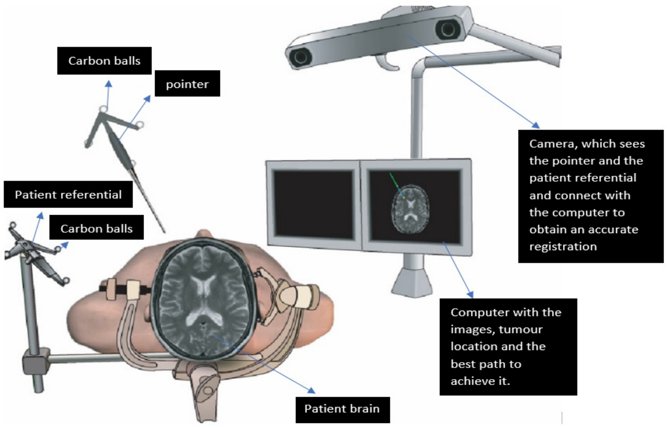


Figure 1: Neuronavigation components[8]

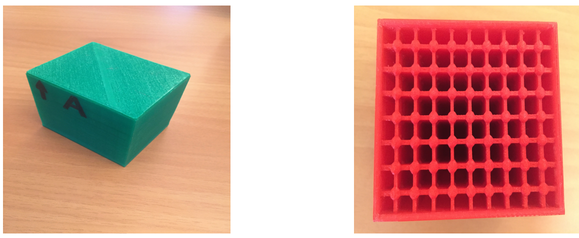
In order to use the image guided surgery system in the operating room, there are several steps the team goes through. The first step is actually before the procedure begin, the CT and MRI images. In fact, CT and MRI work together for the surgical procedure, while MRI gives the brain anatomy and the tumour location, CT gives the skull structure. The fusion between theses images and the patient's head, is called registration and it is the key to identify the tumour location and the path into the patient brain. Registration is done in the surgery room in following way: the patient is lying on a bed where beside their head is located two carbon balls which do not move, the patient [11]. The doctor has a moving pointer, which also contains carbon balls, to point around the patient head. The camera sees the pointer and patient referential and matches those in the 3D data from MRI and CT. Once the registration is obtained, the doctor can finally navigate to find the tumour. Nowadays, this process allows making the smallest opening into the bone to allow the doctor to work inside the brain, find, identify the edges of the tumour, check and verify what is possible to remove all of the tumour before finishing the procedure[11].

3. Case-Study

The 3D medical imaging shows inside the human body, allowing to detect in real time the tumour edges and the feasible brain region to reduce the risk of neurological and motor squealer for the patient. In some cases, the acquired images are not reliable enough to guide on a solid way the conduct with the patient. In this context, the accuracy and the images quality are the key to achieve more complex resection, less traumatic and faster surgeries with less patient recovery time and avoid critical brain areas. However, MRI images present remarkable distortions, that could be a risk for the patient if this distortion mask the real location to be achieved. For that, a Phantom with straight and rigid edges and not deformable internal structure was created to be scanned by CT and MRI machine to compute the MRI distortion. The CT are used as a referential once it does not present any significant distortion.

3.1. The Phantom

The phantom is a grill plastic shell cube filled with gel. It is the referential to analyze the discrepancy between the CT and MRI images. The plastic shell was designed in Solidworks software then printed in 3D. After that it was fulfilled with a hydro gel and closed with a plastic cover also printed in 3D.



(a) Outside the phantom

(b) Inside the Phantom

Figure 2: The Phantom

The phantom are mainly made of two components: the inner gel and plastic. However, some air spot remained inside the cube when it was filled. The inner gel is made of water that contains hydrogen, the main molecule MRI scanner are able to detect/read. The plastic structure is detected/read by the CT scanner. The remaining air are represented by a dark spot in both images and may disturb the image segmentation. The plastic structure is composed of small plastic spheres placed neatly, spaced 10 cm from each other and linked by plastic, making the inner structure similar to a grill. These spheres appear in both 3D images (CT and MRI), with different levels of intensity and contrast due to the different acquisition technology. The 3D images are composed of 2D images that contain a slice of the sphere.

The main objective in creating this phantom is because it excludes many disturbing variables on medical imaging methods, such as motion during the exam, internal changes and human anatomy and the exactly locations of the spheres are known.

4. Implementation

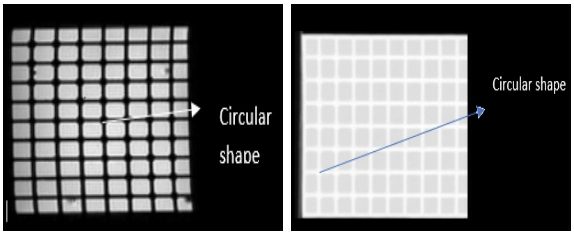
The first step to quantify the MRI distortion is to detect the location of the sphere inside the phantom and shown up in the MRI and CT images. For that, image segmentation and processing methods are used. Then find out the best coordinate system transform, registration, which fit better MRI coordinate system on CT coordinate system. Following this way, the extracted points from MRI and CT images will not have the same location though the object is the same. Then the rigid, affine and non rigid registration are going to be the tools to compute the difference between each point and each image. Since the detection the spheres in the images to the registration are all done with the MATLAB software support.

4.1. Circular Detection

Circle detection task has been a very important topic in image processing field and has a lot of applications in real world[13]. The 3D images from MRI and CT are composed of 155 2D sagital images and of 144 2D Axial images, respectively. The spheres in 2D images appear as circles. So detecting the exactly locations of the circular shapes are very important to achieve an accurate representation of the spheres center point cloud to obtain a realistic relationship between the coordinates of the same points in two different frame, CT and MRI. A typical image of 2D images from MRI and CT is:

These 2D images are from the coronal plane, the best to detect the circular shapes.

The circular Hough transform is the most popular and very well developed approach for solving circle detection problems[13]. Though the circular Hough algorithm is quite good at finding circular shapes with gaps, blur and misshaping. However, with this images, Hough transform was not successful, but a new algorithm based on its concepts was implemented to detect circular shapes on the image



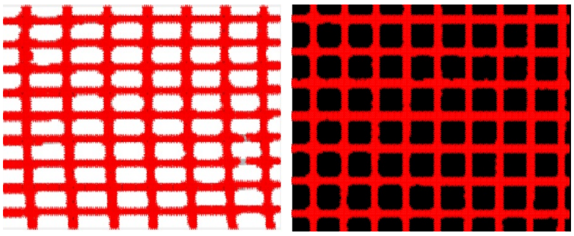
(a) MRI coronal plane (b) CT coronal plane

Figure 3: 2D MRI and CT images with the circular shape

of the phamton.

The new algorithm follows the same steps, even with different intensity and contrast level of the two kind of images, MRI and CT.

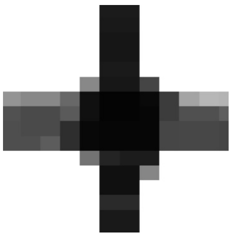
1. Use an appropriate threshold to select only the useful information and select this points.



(a) MRI threshold (b) CT threshold

Figure 4: Threshold selection in red

2. Go to every single selected point and analyze the surrounding pixels and select those which are surrounding by pixels in the threshold.



x	136	23	23	69	x
104	16	4	9	9	60
90	16	4	4	6	60
46	8	6	6	6	70
46	8	6	6	6	70
x	115	36	28	28	x

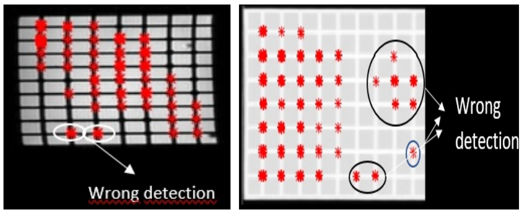
(a) circular shape zoom (b) image pixel description, in pink, pixels surrounding by pixels in the threshold.

Figure 5: Select pixels surrounding by pixels in the threshold

3. Apply restriction to eliminate some odd selection.

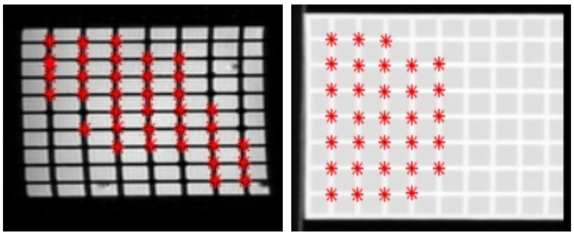
4. Find the center of the circles in x,y axis

5. Find the sphere center in z axis and create a point cloud.



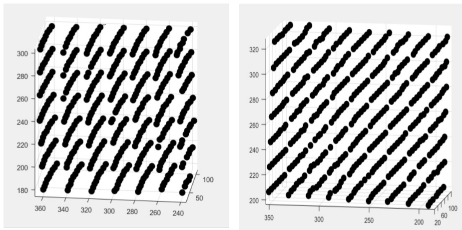
(a) MRI wrong detection (b) CT wrong detection

Figure 6: CT wrong detection



(a) center circle in MRI (b) central circle in CT

Figure 7: circle center detection



(a) MRI point cloud (b) CT point cloud

Figure 8: Point Cloud

Pay attention on the MRI point Cloud, the points in the z axis are not straight as in the CT point cloud. This little curve is the MRI distortion

4.2. Registration

Image registration is often used in medical and satellite imagery to align images from different camera sources[4]. Image registration is the process of aligning two sets of images to establish spatial correspondence between their structures and features.Many times, registration is required to fuse image information from different equipment, detecting image variations produced by different time intervals or under different conditions. In general, the process of image registration involves finding the optimal geometric transformation which maximizes the correspondences across the images[4, 3]. In medicine, the registration process faces many challenges, such as different patient positions, variation in the acquisition of the parameters, different equipment with different physical principles and coordinate systems, and structural variation due to disease or normal temporal variation .There are many processes to determine the best way to do the registration and many methods have been developed based on the image intensities or on the features from the images[3].

After the points extraction, rigid, affine and non rigid registration are done based on the features and on the images intensities to compare their results and accuracy.

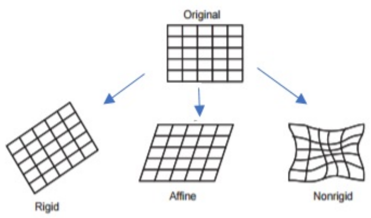


Figure 9: Three registration methods [3]

4.3. Rigid Registration
The rigid registration is a linear coordinate transformation that can be described by 6 degree of freedom (3 rotations + 3 translations)[3]. A rigid registration can be computed by the geometrical features in images, such as points, lines and surfaces, or else by image intensity values. Those based on on features Those based on features determine the transformation T by identifying features such as sets of image points x_A and x_B that correspond to the same physical entity visible in both images and establish a correspondence. In this kind of registration, the first step is to identify and extract the homologous points, called control points or features, and later the correspondence between them is computed using different methods. In the patient, these points can be natural spots, such as bones, or artificial spots introduced into the patients body to help the registration[3]. The other approach, based on image intensity values. works differently. It iteratively determines the image transformation T that optimizes a pixel/voxel similarity measure.

4.3.1 Feature-based Registration:Least-Squares Rigid Motion Using SVD

The rigid registration method based on features uses the least squares rigid motion using SVD method to compute the coordinate matrix. This method optimizes the distance between homologous points.

The main steps of this algorithm is[14]:

1. Compute the weighted centroids of both point sets;

2. Compute the distance between each point to the centroid;

3. Compute the covariance matrix;

4. Compute the singular value decomposition $S = U\Sigma V^T$-available in many programming languages and find the optimal rotation matrix ;

5. Compute the optimal translation (T)

4.4. Affine Registration
An affine registration is a non-linear coordinate transformation described by 12 degree of freedom (DOF): 3 rotation, 3 translation, 3 scaling and 3 shear, 6 DOF more than rigid registration, which is only described by rotation and translation [3]. Those 6 extra DOFs allows to change the geometry of the original image overcoming the limitations in the rigid transformation. The affine transform preserves points, straight lines and planes and does not more preserve lengths and angles.

4.4.1 Intensity-based Algorithm

The intensity based algorithm estimates a geometric transformation that aligns two 2-D or 3-D images [10]. The algorithm operates on the voxel values interactively. It uses the CT and MRI pair of 3D images, a similarity metric, an optimizer, and a transformation type: rigid, similarity, affine or transformation. These parameters are variable according to the image intensity. A similarity metric is a function that measures the resemblance of the intensity values in the two Images[10]. A similarity metric is a function that measures the resemblance of the intensity values in the two images. Different modalities have different intensity distributions. The metric evaluates the accuracy of the registration, returning a scalar value that describes how similar the images are[2]. The optimizer is one of the main components. It determines the best transformation for minimizing or maximizing the similarity metric. The optimizer is responsible for stopping the operation. There are two ways to stop the operation: when it reaches a point of diminishing returns, or when it reaches the specified maximum number of iterations. The parameters of optimizer can be changed as needed. The transformation type determines how to align the misaligned image (called the moving image) with the reference image (called the fixed image). The algorithm starts with the transformation type specified as an input.

4.5. Non Rigid Registration
The registration of biological images poses some challenging obstacles that a simple rigid-body transformation, such as the least squares rigid motion SVD method may not be able to overcome.

[5].It is important to realize that even when the object is rigid, there may be distortions in the imaging process that mean the correct transformation between the coordinate spaces of the source image and of the output image requires additional degrees of freedom. Non-Rigid registration is a family of correspondence functions, capable of expressing essentially arbitrary non-linear relations. The parameter that describes a non-rigid registration has many degrees of freedom: about 2000, describing a free form deformation.

There are many algorithms which compute a non-rigid registration. However, in this thesis, a B spline based on a free-form deformation is used to compute the transformation between the MRI and CT images and the control points extracted from them. The basic idea of this method is to deform an object by manipulating an underlying mesh of control points (grid knots). The resulting deformation controls the shape of the 2D and 3D object and produces a smooth and continuous transformation[1].

4.5.1 B-Splines free form deformation

The main goal of image registration is to relate each point from a fixed image to a point of the moving image and find the optimal transformation which maps the points in the dynamic image into the corresponding points in the reference image. To define a B-spline free form deformation method, firstly an underlying grid is created in the image domain (phi) where it is possible to manipulate the knots and identify the knots locations, being locally controlled. The resulting deformation produces a C2 continuously differentiable transformation [58] that is computed using a cubic B spline transformation function. This function takes as parameters a particular point (pixel or a voxel) $x = (x; y; z)$ and the current grid configuration, and gives the displacement for that point after deformation.The displacement field is composed of x,y, and z components according to the image domain. The vector is decomposed into its components and in each component the cubic B spline is applied to interpolate the points between each grid knot. The location of the points that do not belong to the grid knot is identified by the cubic B spline interpolation in each vector component.

5. Results and Validation

The quantitative validation of the rigid, affine and non rigid registrations was done using the point cloud and the qualitative validation was based in registered images. The coordinate transforms are applied to the MRI points/images to fit the CT points/images that correspond to the fixed referential. The mean, maximum and minimum results are the distance between the CT points and the transformed MRI points, where x,y,z are the 3D coordinates.

5.1. Rigid Registration

5.1.1 Feature-based Registration

The feature based registration was computed using the least square rigid motion using SVD method.

For the MACHINE 1, the mean error between the CT and MRI points is 0.8826 mm. The maximum error is 1.8772 mm and the minimum is 0.0376 mm. The maximum error occurs in the Sagital (x,y) plane with a distortion of 1.875 mm. In Coronal(y,z) plane and Axial plane(x,z) the distortion is 1.5695 mm and 1.41 mm, respectively.

For the MACHINE 2, the mean error between the CT and MRI points is 0.7917 mm. The maximum error is 2.225 mm and the minimum error is 0.0667 mm. The maximum distortion occurs in Sagital plane (x,y) 2.2167 mm. In the coronal(y,z) and Axial(x,z) plane is 1.9876 mm and 1.96 mm, respectively.

The rigid registration results are the real distortion of the MRI, because the rigid registration does not deform the object to compute the coordinate transformation. It only applies rotation and translation matrices which better fit the two sets of data (CT and MRI). Comparing the different rigid registration results acquired by different methods, the feature-based registration got better results because the corresponding points are known and the coordinate transformations are applied to these points and the quantitative validation is done using the same data. However, the image intensity registration does not depend on the extraction accuracy and does not lose image information.

5.1.2 Intensity based Registration

For the MACHINE 1, the mean error between MRI and CT points is 1.6385 mm. The maximum is 4.1369 mm and the minimum is 0.2083mm. The maximum distortion occurs in the Sagital(x,y) plane with a value of 4.04mm. In the coronal(x,z) and Axial(x,y) plane the distortion is 3.5541 mm and 2.63 mm, respectively. The mean value is bigger using the intensity based, once the matrices are computed using the images and are validated with the point clouds.

For the MACHINE2, the mean error is 1.3954 mm. The maximum is 2.8129 mm and the minimum is 0.2587 mm. The maximum occurs in Sagital plane with a value of 2.8 mm. In coronal and Axial plane is 2.57mm and 2.4mm respectively. The images look very similar with the other already displayed, so the point cloud and the planes images are not going to be displayed.

Comparing the mean error from the two different registration, the intensity-based show worst results. Even with not depending on the extraction accuracy and not losing image information, intensity based registration is not so reliable as with the features. Using the features to register the images are much more reliable as it is possible to know exactly the corresponding points between the images.

5.2. Affine Registration

Affine registration is a non-linear transformation: it deforms the image to better fit the data. The matrix R corresponds to the rotation, scaling, and shear parameters, and the matrix t corresponds to the translation.

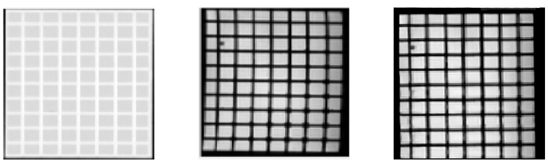
For the MACHINE 1, the mean error between the MRI and CT images is 0.7543 mm and 1.6064 mm for the feature and image intensity based registration respectively. The maximum occurs in the Sagital plane for both king of registration.

For MACHINE 2, the mean error is 0.7960 mm and 1.6957 mm for the feature and image intensity based registration respectively. The maximum is around 4 mm for both method of registration and the minimum is around 1 mm.

5.3. Non Rigid Registration

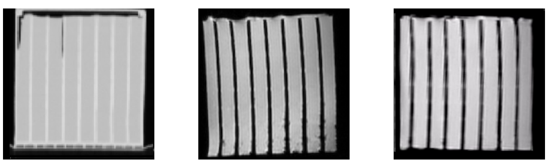
5.3.1 Image Intensity results

Usually image intensity registration leads to many problems with the image intensity correspondence and sometimes yields an inaccurate result. The medical images used in this work are noisier and come from different applications, CT and MRI and it is quite important to get an accurate result.



(a) Fixed Referential-CT (b) Moving Referential-MRI (c) Transformed Referential

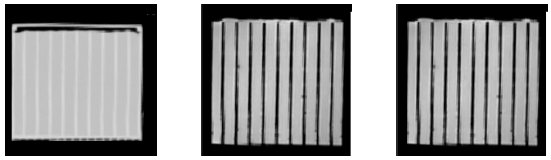
Figure 10: Coronal plane



(a) Fixed Referential (b) Moving Referential-MRI (c) Transformed Referential

Figure 11: Sagital plane

As it is possible to see, the inside columns in the transformed image became straighter than the moving image. In the coronal plane, the columns and



(a) Fixed Referential (b) Original Image Moving Referential-MRI (c) Transformed Referential

Figure 12: Axial plane

lines inside the cube are visually straighter, which means that the MRI deformation was reduced. In the sagital plane, the columns still curve, but it has been barely attenuated. The coronal plane gets better results once the algorithm was computed in this plane and then applied to the whole 3D image domain. As was expected, in the axial plane, it is not possible to observe any visual difference since the main distortion does not occur in this plane.

The quantitative results for the non rigid registration based on image intensity are:

Table 1: Non rigid Registration based on image intensity : MACHINE 1 and MACHINE 2

MACHINE 1	mean	0.0112
	max	0.0845
	min	0
MACHINE 2	mean	0.0091
	max	0.05
	min	0

5.3.2 Feature based

The displacement field size is a 4D matrix with 132x132x42x3 degrees of freedom. It is impossible to display here their values. The 3D registered MRI point cloud, in blue, and the CT point cloud are in the figure below. Here there are mostly blue points because the blue points overlie the red ones. The mean, maximum and minimum values between the MRI and CT points are in Table below

Table 2: Non Rigid Registration

	mm	MRI 1	MRI 2
Feature based	mean	0.0086	0.0083
	max	0.0781	0.0493

Comparing all the non-rigid registration results, all of them rectify almost perfectly all of the deformation. However, the remaining distortion by the non-rigid registration based on image intensity is bigger than the feature based registration once the displacement field is computed on the same data as it was validated, thus both results are very good.

5.4. Result Analysis

Table 3: Intensity based Registration

	mm	MAC 1	MAC 2
Rigid Registration	mean	1.6385	1.395
	max	4.1369	2.819
Affine Registration	mean	1.6064	1.6957
	max	4.5353	3.4499
Non Rigid Registration	mean	0.0112	0.0091
	max	0.0845	0.05

Table 4: Feature-based Registration

		MAC 1	MAC 2
Rigid Registration	mean	0.8826	0.7917
	max	1.8772	2.225
Affine Registration	mean	0.7543	0.7960
	max	2.0206	2.088
Non Rigid Registration	mean	0.0086	0.0083
	max	0.0781	0.0493

The best results were obtained by the non-rigid registration based on the features. The MRI distortions computed by the intensity based registration are bigger than from the feature registration. Firstly, because the registration uses the image not the features to compute the coordinate transform and the results are validated with the points. Regarding the type of registration, i.e. affine or rigid, there is not much difference in the MRI distortion: the magnitude is very similar. This is due to the fact that the nature of the distortion is warp and not shear. The shear and scaling parameters present in the affine registration are not present in the rigid registration, so if those parameters are not significant, it is better to apply the rigid registration as it deforms neither the points nor the images.

The rigid registration deforms neither the point cloud nor the images. It is a linear transformation of the coordinates, which computes the best rotation and translation matrices to fit the MRI points/images to the CT points/images in the best way. So the rigid registration distortion results show exactly the MRI distortion present in the images. The affine and non-rigid registrations are non-linear coordinate transformations that deform the images and the points to better fit the two referential, rectifying the distortion as much as the degrees of freedom allow. The affine degrees of freedom only allow deforming using shears and scalings. The non-rigid registration has many degrees of freedom, which allows deforming the points and the images in different ways, almost rectifying the MRI distortion. Both machines used to quantify and rectify MRI distortion obtain almost the same results in the different registrations. These results show us that the MRI distortion can be lead for the physical principles.

6. Conclusions

The present work focused on medical images from the registration of MRI and CT image modalities based on features and on image intensities to quantify and rectify MRI distortion. Quantifying and rectifying MRI distortion brings countless advantages for knowing accurately a tumours location. For this, the work can be split into two different main steps. The first main step is to extract the features from the complicated and noisy MRI and CT images using an automatic algorithm. The algorithm implemented detects circular shapes and creates three different point clouds from the two sets of MRI images and CT images. The algorithm obtained a success rate of 93with the first set of MRI images (not detecting 27 spheres); 97set of MRI images (only 8 spheres undetected) and 99values (in millimetres) depended on the axis: 0.5 mm in the x axis and 1 mm in the y axis. The second step is the image registration. The methods, namely, rigid, affine and non-rigid, were applied to the features and to the image intensities. Rigid registration shows the real MRI distortion, as this registration method does not deform Either the features or the images: it only rotates and translates the object. The other methods, namely, the affine and the non-rigid registrations, deform the object in order to better fit the two sets of images. For rigid and affine registration, the average and maximum of distortion are very similar, since the affine method rectifies only the shear and scaling deformation and neither the shear nor the scaling are present in the features or the images. Scaling is not present because it has been already corrected before the registration. For these methods, the computed coordinate transformation is the same for each point in the domain. On the other hand, for non-rigid registration, each point has a different coordinate transformation. With rigid and affine registration based on features, the average distortion remained below 1.5 mm and through the sagital plane the distortion increased to a maximum of 2.5 mm. With registration based on image intensity, the average distortion increased to 2 mm with a maximum, also in the sagital plane, of 4.5343 mm. The best registration results are based on features, because the distortion results are quantified using them.

With non-rigid registration based on features, the average distortion is reduced to 0.0086 mm, rectifying almost all the distortion. However, this method yields a displacement field that is not able to deform the 3D image because the displacement field does not have the necessary degrees of freedom to deform the whole image. Finally, the best registration result is obtained through non-rigid registration based on image intensity using the method of free-form deformation with B-splines. The average MRI distortion is 0.0112 mm and the deformation results are visible in the 3D images. Non-rigid registration has many degrees of freedom and it compensates for the distortion in the different planes. It enables a more flexible matching of local details between the two images than does rigid and affine registration. In conclusion, the best approach to rectify MRI distortion is non-rigid registration based on image intensities. The values of the MRI distortion are in the millimeter range. However, this remains significant when we are dealing with complex procedures: a distortion by one millimeter can lead to a fatal mistake.

7. Future work

The presented approaches for quantifying and rectifying MRI distortion can be taken further and improved. Firstly, to verify if the distortion is present in all the MRI machines and if the distortion comes from the physical principles or from the machine calibration, scanning the same phantom in more MRI machines with different parameters and verifying the magnitude of the distortion would be indicated. As the distortion presentts in a round way, scanning, with the same MRI machines, other phantoms with different shapes, with round edges, and extracting features from tjhem is planned Subsequently applying the coordinate transformations obtained by the rigid, affine, and non-rigid registrations to find out whether the phantom shape changes if the distortions also change. Future developments in image registration should use feature-based methods, as that method showed better results and it is sensitive to the different modalities. Also, feature-based registration reduced the computational cost and allowed determining the correct corresponding points. Trying an hybrid registration that blends a registration based on features and one based on image intensities is a promising avenue for future research. This combination gets the best of each ingredient, and improves the gaps in each individual method. Regarding the accuracy of the algorithm implemented to extract the circular shapes, scanning the same phantom in the same machines again, extracting the circular shapes using the algorithm implemented, then applying the coordinate transformations from the different registration methods and verifying the distortion, is something planned for future investigation. Finally, it would be promising to try to establish a standard regarding how the distortion appears and the parameters responsible for correcting it automatically.

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