HDM: AN HETEROGENEOUS STRUCTURES DEFORMATION MODEL

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1 Introduction

In the last decade Computer Graphics applications in medicine have grown due to the advances of input technologies such as MRA and CT which enable the construction of three-dimensional representation of anatomical structures and their visualization. Volume data representation schemes have been studied, particularly the voxel model for regular input data [Kau90] and, for non-regular data sets, the tetrahedral cell model [CFM+94] along with more compact representations such as octrees [WG92], multitetras, and frequency domain representations, as wavelettes [Mur93]. Major attention has been paid to the problems of the accuracy of the representations, their memory requirements and their adequacy for the visualization. However, these models are generally static: the representation of the temporal evolution of volume data, particularly their deformation has been less addressed.

Different medical applications need the simulation of deformations of the shape of anatomical structures. As an example, in oncological studies, it is often necessary to simulate or to predict the growth of a tumor which may produce the deformation of the surrounding structures and therefore provoke secondary pathologies. A particular case of this, are the embolisms caused by the stenosis of a cerebral vessel due to the pressure of a brain tumor. Other examples of deformations in medicine are those produced by external forces, such as the bisturi pressure.

In this paper, the problem of the deformation of heterogeneous volume data sets is analyzed. A general method is proposed, enabling the simultaneous deformation of both the interior and the shape of various imbricated structures. The method supports different deformation models according to the particular hypothesis of each structure elastic behavior. The paper is structured into three sections: first the previous work is reviewed, next the proposed method is described and finally some simulation examples are discussed before the conclusions.

2 Background

2.1 Previous work

Shape deformation consists in the modification of either the geometry or the topology of a surface. There are two main deformation behaviors: elastic ones, which are associated to geometrical modifications, and non-elastic ones, which correspond to structural changes such as fractures. Most of the deformation literature has focused at shapes represented explicitly either as polygonal meshes or smooth sculptured surfaces, although, recently, some papers address the deformation of discrete, voxel-based, surfaces.

Surface deformation has been mainly used for the following purposes:

- Simulation of physically realistic deformations such as those produced by objects collisions and material heating and fusion. [TW88], [KT91].
- Smooth surfaces interactive design by successive deformations of an initial rough shape [SP86], [Coq90], [HHK92] and sculpturing of discrete surfaces in binary voxel models ([GH91],[WK95])
- Animation of non-rigid structures, such as articulated or legged figures [GVP91] and blobby models ([WMW86]).
- Morphing between two shapes [KCP92], [BN92], [CSB95] and morphing between two volume data sets [Hug92], [LGL95].
- Segmentation of regions of interest in 2D images and 3D volume data by successive deformations of a temptative curve (snake) [KWT88] or a polyhedral approximation [MBL+91]([GW92]) of the shape of the region.
- Matching between two geometrical or volume models in order, for instance, to compare the anatomy of a patient with a previously created representation of an anatomical atlas model ([BK89]) or to match two different modalities of registration [Dav97]
- Reconstruction of a binary discrete 3D model from a set of semi-transparent image projections of it ([Mur91]).

Shape deformation models fall into three main categories: kinematic, dynamic and modular models. The kinematic models enable to compute the deformed model on the basis of geometrical information only, typically by interpolating new positions using a user-specified subset of displacements. Two main different kinematic approaches have been published: the application of non-linear geometrical transformations such as bending, tapering and twisting ([Bar84], [CR94] and Free-Form-Deformations ([SP86]) along with their numerous extensions ([Coq90], [CJ91], [HHK92]). Dynamical models simulate physically realistic behaviors and model the deformations produced by forces and torques according to physical laws of elasticity and materials resistance [TF88]. Modular models, also called *layered* models, are based on a multi-level representation of the objects: a first simplified layer, to which dynamical deformations can be applied, and a second layer, composed by the surface or skin of the structure tied to the kernel layer according to kinematic constraints [GVP91].

2.2 Statement of the problem

Most of the previously cited deformation models are *surface-based*, i.e. they assume that the objects are empty and they do not address the impact of the surface deformation on their internal structure [CEO+93]). This hypothesis, valid in numerous cases, is no longer acceptable in many medical applications where the real behavior of the objects should be simulated. Several attempts have been done to model the 3-D volumetric nature of the objects such as [CEO+93] and [BNC96] which generalize to 3D the elastic deformation model, using a 3D finite element mesh, in order to better simulate surgical cuts. However, these models assume that the interior of the objects is homogeneous or, at least, that the property values in their interior remains unmodified under deformation. In addition, the deformation is applied to a single 3D anatomical structure model extracted from medical images.

Nevertheless, anatomical regions are composed of various homogeneous and heterogeneous imbricated structures. The surface of an anatomical structure cannot always be deformed aside from the surrounding regions, because its deformation may provoke the deformation of the external and the internal structures as well. Figure 1 illustrates this idea. In figure 1.a a model composed of two-regions is represented schematically with two different fill-area patterns. The surface of the internal region has been identified. In figure 1.b, this surface has been deformed independently from the property values of the surrounding volume. The effect produced is that the boundary no longer encloses the circle-pattern region. Figure 1.c illustrates the desirable result, in which the volume properties have been modified in accordance to the surface displacement.

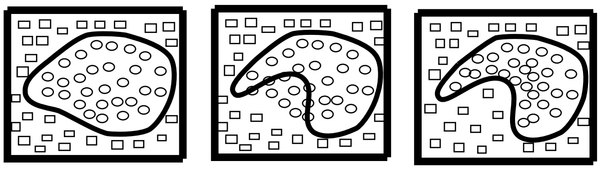

Figure 1: Interrelationship between volume and surface deformations

Herein a general framework for the deformation of volume models is proposed. Its main feature is that it is *hybrid*: it enables deformations of both the surfaces and the volume. This is accomplished by propagating the surfaces displacements to their internal and to their surrounding volume and therefore, to the other embedded structure boundaries. This general framework is suitable for any deformation model, kinematic as well as dynamic. Herein however, a specific development is analyzed, based on a kinematic model with constraints. This model will be referred as HDM *(Hybrid Deformation Model)* in the rest of the paper.

3 The Hybrid Deformation Model

3.1 General framework

The pipeline of the proposed framework is illustrated in figure 2. The representation model is a grey-level, heterogeneous voxel model embedding different regions corresponding to various anatomical structures: organs, bones etc. This model is obtained by registration of data using medical devices such as CTs and MRs and by applying filtering and segmentation processes which eventually remove the noise of the input slices and enable the identification of the anatomical regions. The boundary surfaces of the structures are not represented explicitly but they can be extracted from the voxel model either using a marching cubes algorithm or by contour extraction and tiling between successive contours. The application receives as an input a set of displacements of points belonging to the surface of one or more anatomical structures interior to the voxel model. These input points can be obtained in different ways: they can be, for instance, the coordinates of an electronical scalpel pressing the surface of an organ, or they can be interactively sampled on the surface of a tumor contiguous to anatomical structures, whose deformation is being studied. With these input data and the original model, the application computes the shape deformation and the resulting modification of the properties of the voxels inside and outside the structure. The output of the application is a new grey-level voxel model embedding the deformed regions. This new model can be manipulated as the original one, i.e., it can be visualized, the deformed surfaces of the inner regions can be extracted from it, etc.

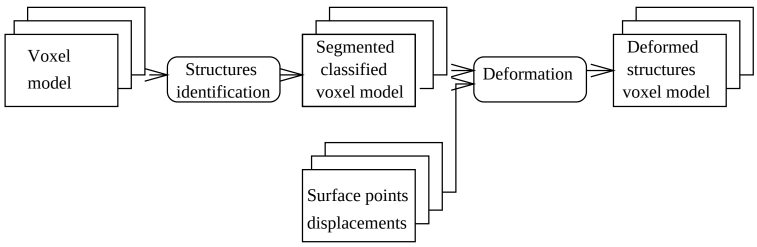

Figure 2: Pipeline of the proposed method

3.2 Description of the method

3.2.1 The data

As mentioned above, the discrete representation of the volumetric model before the deformation, V, is a voxel model such that:

$$V = \{v_{ijk} | prop(v_{ijk}) = \rho_{ijk}\} \quad (1)$$

where the voxel v_{ijk} is characterized by its property values, for instance, its density ρ_{ijk}.

The different regions, or anatomical structures, inside the voxel model have been previously segmented in such a way that each region has an unambiguous property value range. [1]. The region surface pass through the boundary voxels characterized by a non-homogeneous neighborhood. In addition to the classical transfer functions which associate to the different property ranges opacity and color values, new transfer functions have been designed which provide elastical properties for each range. These functions are empirical, based on the physicians knowledge.

The input data of the deformation, D, are pairs of homologous points of the region surfaces before and after the deformation:

$$D = \{(p_1, p'_1), ..., (p_n, p'_n) | \, \forall i = 1..n \; p_i \in S \;\; p'_i \in S' \quad p'_i = Deform(p_i)\} \quad (2)$$

where S is the initial surface structure and S' is the deformed surface structure.

3.2.2 The deformation

The proposed method is composed of three consecutive steps, called *Identification*, *Deformation* and *Restructuring*.

At the first stage, the surface of interest is identified inside the voxel model V. From the input data D of the desired deformations and, applying a surface deformation technique, the deformation stage deforms the whole voxel model, obtaining a non-regular lattice model. The restructuring step computes a regular voxel model equivalent to the non-regular lattice model.

Each step may be performed according different strategies. Next a particular implementation of this pipeline is described. However, any other surface identification and surface deformation could have been applied as well.

Identification As mentioned above, the HDM deforms both the surface and the volume. Therefore, both information must be represented simultaneously and point one to each other. Thus, the requirement of the surface identification is to create a polygonal model of the surface, to which a surface deformation model could be applied, while preserving information of the voxels to which the surface faces belong.

The technique most used to extract a surface from a voxel model, is the *Marching Cubes* technique [LC87], which gives an approximation of the surface as a set of up to three polygons inside each voxel, using trilinear interpolations of the property values to compute the surface vertices. The marching-cubes surface model is complex, made of many small faces, and it looses the information of the voxels to which the vertices belong, unless specific data structures are designed to keep this information.

Herein, a simpler approximation of the surface is used: the cuberille model [UG93] which is composed of the boundary faces of the boundary voxels. Before

[1] In MRA data, representing vascular information, this range segmentation is not always feasible, as vascular structures do not correspond to specific ranges but to local maxima. In this case, the volume model is constructed after the segmentation using labelling property values instead of the original data.

the deformation, this surface is composed by isothetic faces which are parallel to the coordinate planes. After the deformation, these faces are obviously no longer isothetic. Although the surface representation is rough, it has a lower memory requirement than the marching cubes model, because it needs to store only one vertex per voxel. Boundary voxels and their boundary faces can be computed on the fly.

Deformation

The deformation stage itself performs two related processes: the computation of the displacements of the vertices of the voxels under deformation *surface-deformation* and the computation of the new property values of the deformed voxels *volume deformation*.

The surface-deformation implemented herein is an extended kinematic model based on the *Free-Form deformation (FFD)* [SP86] method. It consists of three steps:

1.- Creation of parallelepiped regular lattice enclosing the cuberille surface of the whole anatomical structure. A local coordinate system is associated to this lattice, having a vertex of the lattice P_0 as the new origin and being the main directions S, T, U parallel to the lattice.

The vertices V_{ijk} of the lattice cells (*control points*) before the deformation are:

$$V_{ijk} = P_0 + \frac{i}{l}\mathbf{S} + \frac{j}{m}\mathbf{T} + \frac{k}{n}\mathbf{U} \tag{3}$$

where l, m, n are the number of subdivisions of the mesh in each direction.

2.- Specification of the deformation in terms of displacements of the control points of the lattice. As the input deformations D are points of one or more cuberille surfaces being deformed, the displacements of the control points should first be computed. This problem has been addressed in [HHK92]. It requires the computation of the pseudo-inverse of a matrix to derive the displacements of the control points that minimize the error between the specified displacements and the actual deformation using a squared difference error metrics. As pointed out in [LWCS96], the pseudo-inverse matrix computational cost is very high when the number of points that should be moved is large. Therefore, the proposed method uses the approach of [LWCS96] which produces similar results to [HHK92] without calculating the pseudo-inverse matrix.

The interface of the points deformation specification restricts the range of the allowable displacements and guarantees that the FFD lattice is still structured and that its cells do not auto-intersect.

In addition, some simple dynamic constraints can be taken into account, preventing the vertices of some structures interior to the deformed cuberille surfaces from being modified. This enables, for instance, to deform structures such as the skin while keeping the bone unmodified. In order to keep rigid structures, the closest lattice vertices enclosing them are fixed and again the interface prevents the specified deformations to break the regular structure of the lattice.

3.- Computation of the displacements of the vertices of the cuberille surface and of the vertices of the inner structure voxels. All the vertices are deformed except those belonging to rigid structures. The regular deformation of a vertex is computed according to the formula proposed in [SP86]:

$$P_{ffd} = \sum_{i=0}^{l} \binom{l}{i} (1-s)^{l-i} s^i \left[\sum_{j=0}^{m} \binom{m}{j} (1-t)^{m-j} t^j \left[\sum_{k=0}^{n} \binom{n}{k} (1-u)^{n-k} u^k V_{ijk} \right] \right] \tag{4}$$

The volume deformation consists of computing the property values of the deformed voxels. The volume is considered as an heterogeneous set of various structures which are themselves homogeneous before and after the deformation. Rigid structure voxels remain unmodified, whereas voxels interior to deformed shapes have a constant modified value. In the deformation, the volume of the structures either increases, decreases or remains constant. The density computation is based on the hypothesis that the matter quantity remains constant through deformation and thus, the changes in the density values are proportional to the modification of the volumes. Finally, as a result of the deformation stage a non-regular deformed lattice model is obtained.

Restructuring

At this point of the pipeline, the original voxel model has been deformed and it is now non-regular although it is still structured and topologically equivalent to its initial shape. The restructuring stage allows to regularize it, while keeping the deformations of the inner structures. This stage is thus essentially equivalent to a re-voxelization [Han90], although, by opposite to other re-voxelizations produced by affine transformations such as rotations of the volume it has to deal with a non-affine transformation.

The equivalent regular voxel model is computed as the same resolution as the original one. However it may be computed at any resolution as well. The restructuring stage consists thus in a scanning the regular voxel model and computing for each voxel which voxels of the deformed model have a non-zero intersection with it.

It should be noticed that a voxel of the lattice model can correspond to one o more voxels in the voxel model. If the correspondence is unique, the regular voxel property is simply set to the deformed voxel property. However if more than one deformed voxel intersects the regular one, their respective property value are weighted and accumulated (see figure 3).

4 Simulations and results

Color Plates 1 to 8 show the results of a 2D prototype simulation of the deformation model. Being a 2D, it would be more proper to talk in terms of pixels

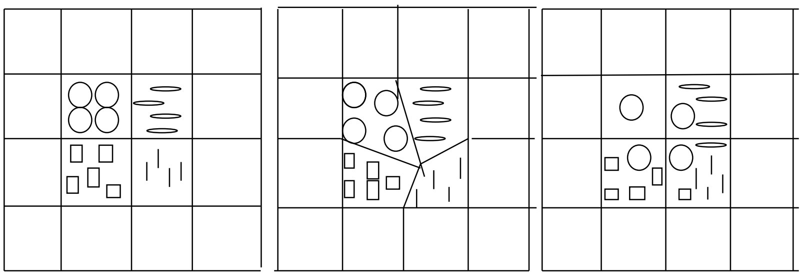

Figure 3: Restructuring

rather than voxels, however, for coherence with the rest of the paper, the term voxel has been preferred.

In Color Plate 1 the original image data of a CT scan of a head are visualized. Two cuberille surfaces are identified in the image: the external surface of the brain and the surface of a tumor, depicted in blue and red respectively in Color Plate 2. The elastic behavior of both surfaces is considered identical.

The FFD net computed as the bounding box of the whole object is shown in Color Plate 3. From specified values of displacements of several points of the brain surface, the deformation of the control vertices of the FFD are computed and represented in Color Plate 4.

Color Plate 5 shows the results of the deformation on the volume. In order to allow a better understanding of the image, *macro-pixels* of 10x10 are represented. It can be observed that the original voxel model is no longer isothetic. The deformed cells behave as closed compartments which *drag* the matter inside them in their deformation. The color of the cells changes according to the modification of the density value inside the cell. The new density is computed as the previous value of density multiplied by the ration between the previous voxel area and the new voxel area. The deformation has been applied only at the voxels which are inside and on the cuberille surface of the brain. Being inside the brain, the tumor is also deformed. Color Plate 6 shows the volume once the restructuring step has been performed. The general aspect of the image is quite similar to Color Plate 5. However the voxels are now parallel to the coordinate axis. The voxels which fall completely inside a deformed structure are considered homogeneous and therefore they are not modified. The voxels which exhibit differences with Color Plate 5 are those that intersect the surface, because their value is computed as a weighted average of the deformed voxels which cover them.

Finally Color Plates 7 and 8 show a more complex deformation.

5 Conclusions and future trends

A general framework for the 3D deformation of multiple structures represented implicitly in a volumetric voxel model has been proposed. The model is hybrid in the sense that it enables the deformation of a surface inside the volume model and the modification of the internal property values as a result of the compression or expansion of the deformed surface. The surrounding volume is also modified in relation to the deformation.

A first kinematic prototype implementation of the model on 2D images, based on the use of FFD has been described. The results of the simulations are encouraging and make it necessary to implement the three-dimensional version of the model. This implementation is currently being done.

The integration of dynamic constraints to the model is another research line under progress. Up to know only homogeneous and rigid behavior are allowed. The use of the true elastic properties should be enabled. Non-homogeneous propagation of the deformation though the volumes properties should also be studied.

Finally, a future extension of the method is its application at different levels of resolution of the voxel structure, enabling higher precision in zones of interests and coarse deformation in areas of less relevance.

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