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# Toward the Simulation of the Strain of Female Pelvic Organs

M - E. Bellemare, N. Pirro, L. Marsac, O. Durieux.

**Abstract**— Surgical simulators have undergone a significant development, especially since the rise of mini-invasive surgery. The main simulators of digestive surgery have been developed for solid organs such as the liver and spleen. Studies relating to soft tissues like the pelvic organs are rare. The aim of this work was to create a model of female pelvis that will be used as a basis for the development of a patient specific pelvic simulator. A dynamic MRI acquisition is used to evaluate and to set the simulation model.

## I. INTRODUCTION

SURGICAL simulators have seen a significant development, in particular since the rise of minimally invasive surgery by laparoscopy. The main simulators of digestive surgery have been developed for solid organs such as the liver and spleen [1]. Studies relating to soft tissues, like the pelvic organs, remain rare. The aim of this study is to generate a model of the female pelvis, to report the constraints and the difficulties of this modeling, and to outline the prospects of such a model. The validation of a physiological model is the first step before the elaboration of a pathological model for a specific patient suffering from pelvic floor diseases. The anatomical abnormalities of such a model could be corrected by simulating the modifications provided by future surgery thus allowing a functional evaluation of the possible surgical procedure.

This paper presents a first stage toward this goal, the simulation of the behavior of female pelvic organs under strain condition. From a volume MRI acquisition a patient specific anatomical model is built and associated with a physical model to perform physiological simulation. A dynamic MRI acquisition, available with the radiological assessment, is used to perform evaluation and model tuning.

## II. CLINICAL RELEVANCE

There has been a recent expansion of three-dimensional imaging of pelvic structures. It has been used to assess the relationship between the pelvic organs, to quantify and to

identify the morphological changes of the organs in different grades of prolapse [2]. Dynamic and animated models simulating the displacement of the pelvic organs could allow a dynamic analysis of their morphology and their relationship during straining. This could lead to a better understanding of the pathophysiology of pelvic floor disorders. Indeed, although significant research has been carried out, the pelviperineal physiology and the anatomical basis of pelvic floor diseases remain unclear and their management has not been optimized. Although various surgical procedures have been advocated for the treatment of pelvic floor diseases, recurrence is frequent and may be the result of failure to identify the causative lesion or to provide optimal treatment [3].

Various models have recently been used to train health professionals in the realm of clinically applicable tasks[4]. Yet, these models are usually interactive animation techniques simulating mainly rigid bodies. The development of patient-specific simulators could allow repeated practice of really complex maneuvers taking into account the anatomical variations and physiological properties of each individual patient [5].

The simulation considered here concerns mainly the physiological movements of the main organs of the pelvic area. It aims more at surgery planning than at practice.

## III. BUILDING THE GEOMETRIC MODEL

### A. Segmentation Stage

The geometrical model that we generated is carried out starting from MRI acquisition of one patient and therefore is a patient-specific model. MRI modality is used, as it is best suited for soft tissue discrimination, it provides both a volume image and a dynamic sequence of the organs at rest and under strain condition. Fully automatic segmentation of such data remains an open issue as it relies in a context of free-form organs with little contrast. We use a semi-automatic method that involves an expert in the process. The segmentation software is based on an open source and free software, ITK-SNAP [6], which we enhanced to fit the specific data. The segmentation process is based on active contours computed with a level set implementation [7]. The active contour is a surface  $C(u,v,t)$  of which evolution is driven by a differential equation (1):

$$\frac{\partial C}{\partial t}(u, v, t) = F \cdot \vec{N} \quad (1)$$

where:  $F = \alpha \eta + \beta \kappa$ ;  $\beta \cdot \kappa$  (Internal force) is related to the Gaussian curvature; and  $\alpha \cdot \eta$  (External force) is linked to organs borders.

The original software method was modified and completed with a set of new tools. One or several volumetric

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germs are placed manually within the areas to be segmented, the boundaries of which are defined by a manual thresholding of the volume limiting the extent of the active contour evolution. This initialization determines the surface  $C(u,v,0)$  as the union of the germs surfaces. The growth of the germs depends on their expansion weight and bending weight  $(\alpha,\beta)$  that are modified according to the thresholding and heterogeneity of the organs. The possibility for the user to modify the weighting map was developed so as to achieve the segmentation of heterogeneous areas. An automatic stopping criterion based on the variation of the volume  $V$  of the volumetric germ was also developed. Volume variations are computed every 10 iterations ( $i$ ). Stopping is automatic when the ratio  $V_{i+10}/V_i$  is smaller than 1.01.

At the end of the segmentation process, geometric irregularities are automatically corrected through the modification of the parameters of the volumetric germ that evolved with a prevailing bending weight (a lower  $\alpha$  value) during 40 iterations (Fig. 1).

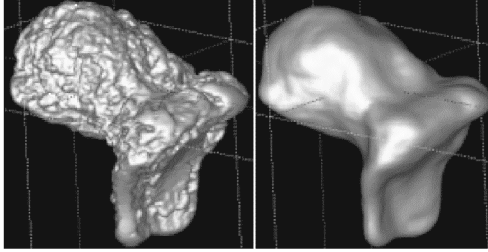


Fig. 1. Smoothness improvement by using a local and geometric criterion. Segmentation results without smoothing on the left, and after smoothing on the right.

From these images, 3D surface meshes of the pelvic viscera and bones are reconstructed with a marching-cubes algorithm and a volume-rendering method provides interactive visualization. In order to reduce the computing time of the following deformation step, the number of triangles forming the surface of segmented organs is reduced without too much geometrical simplification. This reduction is achieved by using a decimation algorithm applied to locally coplanar triangles and by the reduction of the number of faces of areas with a weak Gaussian curvature.

### B. Tetrahedrization

Real world organs may be hollow but they are not empty. Here, the pelvic organs are considered as volumes in order to take into account their volumetric properties. To create a volumetric model, the 3D surface model is filled with tetrahedra thanks to a Delaunay method [8]. Volume tetrahedrization is performed using tetrahedron quality checking based on the minimum volume of the tetrahedra. Typically, from 1500 to 2000 surface points we obtain 6800 to 7800 tetrahedra based volumes for the main three organs considered.

The areas which have too homogeneous grey level or which lie outside the acquisition area, remain quite hard to be segmented, and therefore are represented by simple 3D geometric models. These models are positioned by using landmarks within the volume image. Thus, the whole geometric model obtained serves as the support of the mechanical model (Fig. 2).

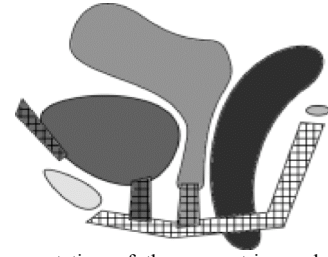


Fig. 2. A slice representation of the geometric model completed with simpler tetrahedra (squared areas).

## IV. PHYSICAL MODEL

### A. Mass spring network

The geometric model was integrated within a model of physical deformation to simulate the physiological displacement of the pelvic organs. The method of mass-spring network is a physically based technique that has been widely and effectively used to model soft tissues. The physics of this model is simple and well understood. The mass-spring network has a simple structure and it enables many operations such as large deformations and topology changes. This physical model consists of a lattice structure of mass points connected by elastic links. The mass-spring network is mapped onto the geometrical mesh, i.e. the masses are the vertices and the springs are the edges of the mesh (Fig. 3).

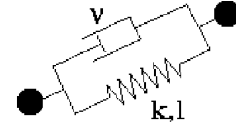


Fig. 3. The mass-spring model (Voigt model) between two vertices.

The model relies on a classical equation (2) : given a vertex  $i$ , its position vector  $x$ , its mass  $m$ , the links forces  $F_{i \rightarrow j}$ , collision forces  $F_{ci}$ , and external forces  $F_{ei}$ :

$$m\ddot{x} = \sum_{j \in J} \vec{F}_{i \rightarrow j} + \vec{F}_{ci} + \vec{F}_{ei} \quad (2)$$

Two neighbor vertices,  $i$  and  $j$ , create a link force depending on: their distance,  $d$ ; the spring stiffness parameter  $k$ ; the spring resting length  $l$ ; the speed difference between them  $\vec{V}_{ij}$ ; and the damping parameter  $v$ ; and leading to equation (3):

$$\vec{F}_{j \rightarrow i} = k \cdot (d - l) \frac{\vec{U}_{ij}}{d} + v \cdot \vec{V}_{ij} \quad (3)$$

Once these equations are set, inter organs interactions can be computed.

### B. Collision

Within this model some tissues are static like bones and some muscles. Interaction between tissues may create collision or links. The computation of colliding forces is based on penalty principle [9]. If two objects  $A$  and  $B$  are overlapping, for every overlapping tetrahedrons pair  $T_a$  and  $T_b$ , where  $T_a$  is from  $A$  and  $T_b$  from  $B$ , repulsive forces are applied. Each force intensity (4) depends on: speed difference  $\vec{V}_{ab}$ , between  $A$  and  $B$ ; their intersecting volume  $V_{int_{ab}}$ ; and the weights  $\mu$  and  $\lambda$ . In order to simplify the

computation, the tetrahedra intersection is based on their bounding boxes

$$\vec{F}_{b \rightarrow a} = \mu \cdot \vec{V}_{ab} + \lambda \cdot V_{\text{int}_{ab}} \frac{\vec{V}_{ab}}{\|\vec{V}_{ab}\|} \quad (4)$$

Thus, a tetrahedron  $T_a$  is subject to the colliding force (5) caused by every  $T_b$  tetrahedrons:

$$\vec{F}_{c_i} = \sum \vec{F}_{b \rightarrow a} \quad (5)$$

### C. Simulation

The diagram (Fig. 4) depicts the computation loop used to determine the vertices positions for each instant  $t=k.dt$ , where  $dt$  is a fixed time step. Each iteration implies first the computation of the different forces and then the equations integration. The internal forces are computed independently for each non-osseous organ in order to create smaller systems. The collision computation is the longest step because of the high number of tetrahedra and the complexity of the algorithm ( $O(n^2)$ ). The computation of external forces is fast as a few points are linked and the strain forces are known. Equation integration is done with a classical method based on a 4<sup>th</sup> order Runge-Kutta algorithm implemented in the GSL library [11].

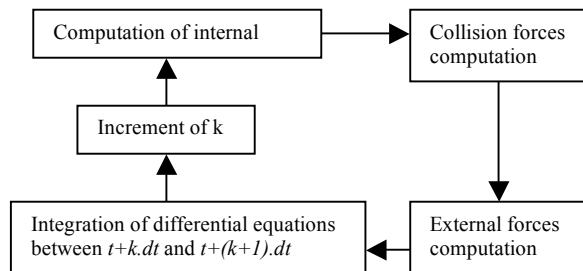


Fig. 4. The simulation loop.

The external forces concern the abdominal force and the diaphragmatic force. These forces are the cause of the behavior that should be simulated. The abdominal force is applied on the points located on the anterior part of the uterus and the bladder whereas the diaphragmatic force is applied on the points located upon the superior area of these organs. The force intensity evolves following a time profile that determines the simulation behavior.

The mechanical parameters of the organs are not obviously defined. For each organ we made the approximation of isotropy that implies all vertices have the same mass, the same spring and damping parameters. Spring resting length corresponds to the length of each edge of the geometric model at the beginning of the simulation. The physical and mechanical parameters of the pelvic organs were unknown, they have been defined by an empirical approach; uterus being the most rigid non-osseous organ.

## V. DISCUSSION

### A. Results

A complete model of the female pelvis was generated from MRI. The MRI acquisition of the median pelvic organs required for segmentation was carried out as part of the

clinical activity of the radiologist without causing great inconvenience. The whole pelvic acquisition (static and dynamic), lasted approximately 20 minutes and remains compatible with routine procedure. The complete segmentation and correction of geometry irregularities required 5 hours. The segmentation method includes a semi-automatic procedure that allows the physician to supervise the process. The learning curve for the segmentation software mastering is relatively short. The segmentation of the bladder and bones was easy whereas rectum and uterus segmentations were more difficult owing to the organs heterogeneity. Two hours have been necessary to model and to position in three-dimensions the regions which could not be segmented. Produced thanks to a pretty good segmentation stage, the geometric model integrates all the morphological characteristics of the patient and allows the 3D visualization of one or several organs simultaneously.

The volume tetrahedrization required 2 hours. The calculations of the model deformations lasted up to 5 hours using a Linux 2GHz Pentium IV workstation.

More than a hundred simulations were carried out modifying one by one the physical parameters of the model to obtain a satisfactory result. As expected, the results were excellent for the whole model at rest with a perfect agreement between the images of the model and the MR imaging at rest. At the end of a simulation corresponding to maximum strain, agreement between the MR imaging and the images of the model was satisfactory for the pelvic organs and excellent for the perineal floor. In general, the volumetric variations during simulation were inferior to 15% as reported in the figure 5.

Volumetric conservation is one of the limits of the method that requires the development of new algorithms such as in [11]. The volume variations of the bladder in our study were quite large. This problem could explain the absence of perfect agreement between displacements of our model and those observed by MRI. A more precise characterization of the physical parameters of the various organs associated with a better volumetric conservation could allow an even more accurate simulation of real displacements.

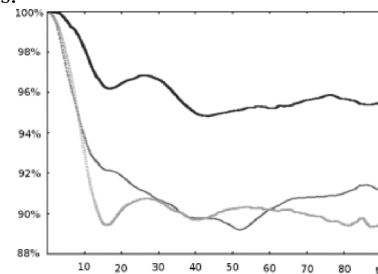


Fig. 5. Volumetric variations of the three main organs during a simulation. The rectum has the smaller variation, uterus-vagina has the fastest variation, and the bladder relies between both.

### B. Validation

The characterization of the physical parameters of the pelvic organs is a preliminary stage essential to the development of a simulator of pelvic floor disorders correction. Very few studies reported the elastic properties of tissues encountered during surgical procedures. Yamada

[12] performed comprehensive mechanical studies on most organs but they were limited to cadaver studies and focused on ultimate strain limits of tissues rather than the elastic behavior within the normal physiologic range. It is acknowledged that a major issue with mass-spring based models is the assessment of physically meaningful model parameters [13]. In our study the parameters were defined by means of an empirical method. Tuning the spring constants of the lattice could be done automatically with simulated annealing or genetic algorithm [14], but we focused first on the feasibility of the study based on the use of dynamic MRI as ground truth.

The validation of our model is achieved by matching a medial sagittal slice built from the model and the MR images acquired in the matching plane in dynamic mode (Fig. 6). This validation can be completed by a quantitative evaluation by measuring 3D coordinates of specific landmarks from the MR images and from the model. The comparison of the movement of these points will permit a better evaluation, both static and dynamic. It will also allow the automatic measuring of several anatomical parameters. In this study, organ displacements were assessed neither in the frontal plane nor in the transverse plane. Only some recent works have estimated these displacements during straining [15] and they appear to be limited.



Fig. 6. Comparison between the result of the simulation and the sagittal MRI dynamic acquisition. White dots correspond to a slice of the 3D model at maximum strain. The model slice was manually matched to the MRI slice.

This preliminary study allowed the development of one specific model, static and dynamic, achieved from the MR imaging of one individual patient. The next steps of our developments involve a quantitative validation methodology based on measurement and matching. Beyond physiology understanding, the validation of the physical parameters could be used as a basis for a simulator of pelvic surgery.

## VI. CONCLUSION

Pelvic physiology and pathology remain difficult problems to model because of the presence of several mobile and deformable organs. In this study the development of a specific model of the female pelvis that simulates straining was achieved although the detection and management of

collisions involving deformable organs still remains time consuming. This work was not focused on speed but rather on the ability to benefit from the “ground truth” provided by the dynamic MRI acquisition in order to tune the model. The precision of the organ models is quite high after segmentation. By reducing this accuracy faster simulations could be performed, depending on the purpose of the application. In addition the physical model could be refined at a computational cost [16-17]. Making this model was the indispensable preliminary study for developing a patient-specific simulator of pelvic floor diseases. The main results are encouraging and the selection of consistent landmarks should provide the set of parameters that will produce a more realistic simulation. A study aimed at the quantitative validation of organs behaviors is currently a work in progress.

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