Registration of a Validated Mechanical Atlas of Middle Ear for Surgical Simulation

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Abstract. This paper is centered on the development of a new training and rehearsal simulation system for middle ear surgery. First, we have developed and validated a mechanical atlas based on finite element method of the human middle ear. The atlas is based on a microMRI. Its mechanical behavior computed in real-time has been successfully validated. In addition, we propose a method for the registration of the mechanical atlas on patient imagery. The simulation can be used for a rehearsal surgery with the geometrical anatomy of a given patient and with mechanical data that are validated. Moreover, this process does not necessitate a complete re-built of the model.

Keywords: Simulation & training systems, Atlases, Head and neck.

1 Introduction

Learning surgery requires a large amount of practice to acquire experience, especially for complex tasks that involve knowledge or delicate gestures achievement. In this context, training simulations are crucial to preserve the patient safety during the learning process. However, to gain interesting experience from simulation, the simulated environment must be highly realistic. In this paper, we present a new approach based on a validated mechanical atlas that is adapted with patient data. We applied this approach for a simulator of middle ear surgery.

Middle ear surgery is a microsurgery of hearing rehabilitation for conductive loss. This surgery is particularly complex due to the high susceptibility to trauma and the sub-millimetric size of the anatomical structures, such as the ossicular chain (Fig. 1). Training on cadavers is the most realistic and appreciate solution but it is expensive and there is a risk of infection [1]. Synthetic models are mainly used for economical and safety reasons [2]. Some virtual simulators exist for ear surgery, like the Visible Ear [3] or the Voxel-Man (Voxel-Man Group, Hamburg, Germany), which are mostly centered on drilling tasks. The main limitations of

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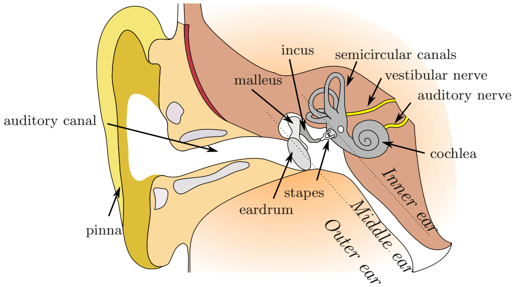


Fig. 1. Anatomy of the human ear. The tympanic membrane and the ossicular chain with the malleus, the incus, and the stapes compose the middle ear.

the simulation are the low number of pathological scenarios. Yet, the anatomy of the middle ear is dissimilar between patients and can lead to different surgical approaches or even to a surgical contraindication.

The objective of this study is the design of a mechanical atlas for the ossicular chain dedicated to the simulation of the middle ear surgery. The mechanical behavior of our atlas is confronted to measurements on human temporal bone specimens. As the atlas is developed for an interactive surgical simulator, the computation efficiency is also investigated. Finally, a registration method of the mechanical atlas to fit patient imagery is also described.

2 Validated Mechanical Atlas Dedicated to Real-Time

This section presents a biomechanical model of the middle ear behavior. The approach is built on finite element methods (FEM) with parameters that were collected from literature. The model is validated for both dynamic small deformations and static large transformations. We then optimize the computation to use the model, without changes, in an interactive real-time simulation context.

Design and Specifications: The three-dimensional geometric model obtained from a micro-Magnetic Resonance Imaging of the middle ear [4] is used to create our atlas. The missing ligaments are manually reconstructed according to the anatomical data using Blender. A volumetric model of the ossicular chain is generated using the Computational Geometry Algorithms Library (CGAL) that allows to create a coherent mesh while defining several domains. We then assign adequate parameters to these different domains (as illustrated in Figures 2(a) and 3(a) with red and blue meshes for respectively, high and low stiffnesses in the domain). The eardrum and the ossicular chain (meshed respectively by 2492 and 4361 tetrahedral deformable elements) are implemented in SOFA, a real-time medical simulation project [5]. Both models use a finite element deformable model (large displacements and rotations) derived from [6], with respectively, membrane and volume elastic energy. The mass is lumped at the

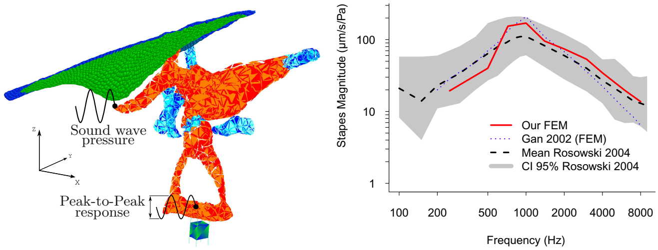
Table 1. Mechanical parameters used in our atlas

Anatomical structures		Density(kg/m^3)	Young modulus (N/m^2)
Eardrum	Pars tensa	1.20 x 10^3 [7]	3.34 x 10^7 [7]
	Pars flaccida	1.20 x 10^3 [7]	1.10 x 10^7 [8]
Malleus	Handle	3.70 x 10^3 [8], [9]	1.41 x 10^{10} [8]
	Head	2.55 x 10^3 [8], [9]	1.41 x 10^{10} [8]
	Neck	4.53 x 10^3 [8], [9]	1.41 x 10^{10} [8]
Incus	Body	2.36 x 10^3 [8], [9]	1.41 x 10^{10} [8]
	Short process	2.26 x 10^3 [8], [9]	1.41 x 10^{10} [8]
	Long process	5.08 x 10^3 [8], [9]	1.41 x 10^{10} [8]
Stapes	-	2.20 x 10^3 [8], [9]	1.41 x 10^{10} [8]
Ligaments	Superior	2.50 x 10^3 [7]	4.90 x 10^4 [8],[10]
	Incudostapedial	2.50 x 10^3 [7]	6.00 x 10^5 [8]
	Incudomalleolar	2.50 x 10^3 [7]	1.41 x 10^{10} [8]
	Malleus anterior	2.50 x 10^3 [7]	2.10 x 10^7 [7]
	Malleus lateral	2.50 x 10^3 [7]	6.70 x 10^4 [10]
	Incus posterior	2.50 x 10^3 [7]	6.50 x 10^5 [7]
Tendons	Tensor tympani	2.50 x 10^3 [7]	2.60 x 10^6 [7]
	Stapedial	2.50 x 10^3 [7]	5.20 x 10^5 [7]

nodes. The tympanic annulus, the stapedial annular ligament, the cochlea, the ligaments and the tendons of the middle ear are used as boundaries of our FEM. The density and the Young modulus parameters of our atlas are chosen according to published data on middle ear mechanics [7], [8], [9], [10], as reported in table 1.

Validation of the atlas: The dynamics of the atlas model is evaluated using the transfer function analysis (TFA) of the ossicular chain. The TFA consists in the measurement of the stapes footplate velocity when a sinusoid pressure is applied to the malleus (Fig. 2(a)). The pressure is set to 0.632 Pa equivalent to 90 dB (SPL) and the frequency ranged from 250 to 8000 Hz like in a physiological condition of hearing. This test cannot be performed in real-time due to the very small time steps required to simulate vibrations at high frequency. The TFA result provides the stapes footplate velocity transfer function (STF) which is increasing for frequency below 1000 Hz and decreasing for higher frequency (Fig. 2(b)). This characteristic shape is also observed by other FEM simulation [8] (but non targeted towards real-time application) and by reported studies on human temporal bone [11]. The large repartition of the TFA reported by Rosowski et al. is explained by the individual anatomical variability of the middle ear components and by the storage condition of the specimens. This test assesses the dynamics of our model at different frequencies (which is linked to the mass and stiffness values) but the deformations are limited to small displacements.

Consequently, we have also conducted the analysis of the umbo displacement under static pressure load (UDSP) to assess the model with large deformation

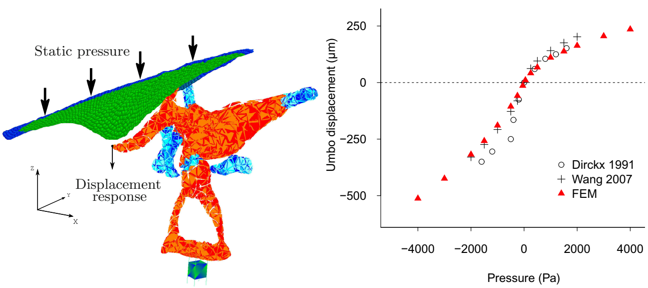


(a) Transfer function method. (b) Transfer function results.

Fig. 2. Evaluation of the atlas in physiological condition using the transfer function analysis. The motion of the stapes footplate is analyzed when a sinusoid pressure is applied to the malleus.

transformation at equilibrium. The UDSP corresponds to the observation of the umbo motion when a static pressure is applied to the eardrum (Fig. 3(a)). We use pressures ranged from -4000 to 4000 Pa which are comparable to pressures involved in a conventional tympanometry diagnostic. As shown in figure 3(b) the results of the simulation of our atlas are close to published observations on human temporal bones [12] or previously published FEM model [13]. Moreover, the ratio of the umbo displacement between negative and positive force is around 1.76 ± 0.31 (n=8) for our atlas, which is similar to the ratio between 1.8 - 1.9 obtained on temporal bones. The analysis of the UDSP allows to assess the realism of the middle ear model when the displacements induced by the deformation are significant compared to the size of the structure (more than 15%). This range of deformation with large displacements is similar to those encountered when simulating a surgery.

Interactive surgical simulation: The goal of this work is to build a training simulation system of a new generation, in which the physics-based models are not over-simplified for real-time performance. Real-time is a needed condition to



(a) Static pressure method. (b) Static pressure results.

Fig. 3. Evaluation of the atlas in surgery condition using the static pressure. The motion of the umbo is analyzed when a static pressure is applied to the eardrum.

allow for the interactivity required for a surgical simulator. This constraint often leads to the choice of plausible models rather than validated physics-based approaches. In this work, we have deliberately chosen to base the real-time simulation on the atlas model that is validated using TFA and UDSP tests. As previously mentioned, the number of elements used in the FEM models is not excessive. We assessed the convergence of the FEM at several resolutions, between the presented model and a model composed of 210689 elements the mean square error using TFA is 2.8 μm/s/Pa, less than 1 dB. Due to the relatively high stiffness/mass ratio, the use of implicit integration is necessary. Still, due to the strong heterogeneities in the model, we had a bad condition number on the system of equations generated by the FEM model. Consequently, iterative solvers, like the conjugate gradient algorithm had difficulties to reach the convergence. Moreover, the factorization cost induced by direct solvers was too heavy for real-time performance. Thus, we have opted for the solution presented in [14], in which a preconditionner is factorized asynchronously in a separate parallel thread (i.e. without additional computation cost in the simulation process). This solution strongly improves the convergence of the conjugate gradient algorithm used in our simulation. The time step of the backward Euler scheme that we use in the simulation is set at 0.04 s. An average of 40 frames per second is observed with no interaction and collision. In the worst case (collision response with more than 15 friction contact points), the number of frame rates falls down to 20Hz. The framework SOFA allows for haptic rendering with Phantom Omni devices (Sensable, Wilmington, MA). The collision response and the haptic algorithm is based on a multithreaded approach [15]. It allows for updating the force feedback at higher rates than the simulation. Using these components of SOFA, we are able to use the atlas model without any changes in the parameters, for a real-time simulation and contact response with surgical instruments.

3 Patient Specific Model

The atlas is developed for training and for rehearsal simulation of the middle ear surgery. Therefore, we implement a registration method to fit our validated mechanical model to the patient image data. The goal is to perform the registration of the atlas (mechanical model, parameters, boundary conditions...) to avoid time-consuming work of segmentation and the re-parametrization. First, a rigid semi-automatic approach is used to place approximately the atlas in the same area and orientation as the patient anatomy. This method is followed by a deformable registration that uses the FEM of the atlas and the implementation of additional spring forces that depends on the intensity of the pixel.

The rigid registration is based on the manual selection of 3 points: The umbo ($\tilde{p_a}$), the incudomalleolar joint ($\tilde{p_b}$) and the incudostapedial joint ($\tilde{p_c}$). These points correspond to distinct landmarks on the ossicular chain that are easily identifiable on the images. These landmarks are also placed on our atlas model, noted p_x (Fig. 4(a)) to allow for a simple rigid registration.

In practice, after the rigid registration, the middle ear components on the images are close to the surface of our atlas. But to enhance the precision of the

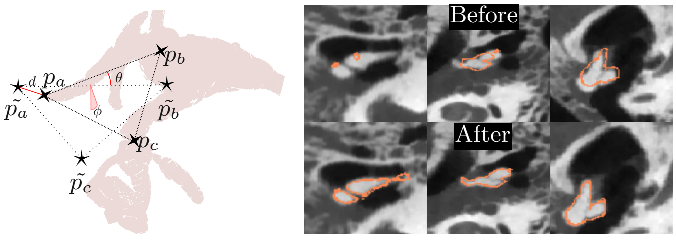


Fig. 4. Registration of our atlas. (a) presents the method of the rigid registration, and (b) the results of the deformable approach, with a representation of the surface of the atlas in orange before (top) and after (down) the deformable registration process.

registration, we perform a deformation based on the variation of the imaging's pixel intensity around the atlas and elastic energy from FEM model. We start from the barycenter B_i, of each triangle i of the atlas surface mesh and we explore the neighborhood along the normal vector n_i inside the mesh ($c = 1$) and outside ($c = 2$) using the equation 1. Where X_i^c is the coordinate of the exploration point along the normal vector, *step* corresponds to the smallest thickness of the imaging slices, and j to the exploration iteration parameter (Fig. 5). The equation 2 computes a pixel intensity ratio between two successive points around the atlas in order to detect an intensity shift meaning that we are at a bone boundary on the patient image. Thus, a border point is detected and associated to each points that compose the triangle i. As a point belongs to several triangles, each point of the surface have several associated border points. Thus, we compute the barycenter ω_k of these associated border points for each point k of the surface mesh. Then a spring is implemented between ω_k and k to deform the mechanical FEM of the atlas. We perform several iterations of the simulation with an update of the border points at each step until an equilibrium position is reached.

Imagery acquisition of four left ears is obtained using a NewTom 5G Cone Beam Computed Tomography (QR SRL, Verona, Italy). The figure 4(b) repre-

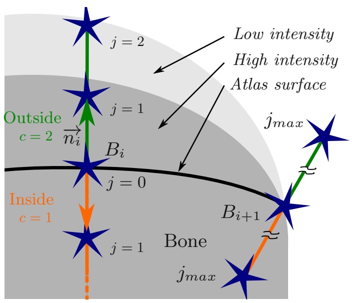


Fig. 5. Detection of the intensity shift around the atlas surface

$$X_i^c(j) = B_i + (-1)^c \cdot n_i \cdot j \cdot step \quad (1)$$

$$R_i^c(j) = \frac{Intensity(X_i^c(j))}{Intensity(X_i^c(j-1))} \quad (2)$$

In figure 5 :

$$R_i^2(j = 2) < 1 - threshold$$

$$R_i^1(j = 2) \approx 1$$

sents the surface of our mechanical atlas over the patient images before and after the deformable process. The results presented are obtained after 300 iterations performed in 110 seconds on a conventional computer. The registration accuracy is compared to manual segmentations using MeshDev [16], a mean error of 0.201 mm is found for our registration algorithm and 0.187 mm between manual segmentations. This error is within the imaging resolution, 0.26 mm, corresponding to the voxel diagonal.

4 Discussion

The objective of the atlas is to supply models for a new surgical tool for training and rehearsal purpose. The simulator is planed to simulate the otosclerosis surgery including stapes footplate drilling or placing the prosthesis to restore the sound transmission. Our simulation, based on a mechanical atlas, provides realistic and validated results for the behavior of the middle ear model. Indeed, results show that our simulation is realistic in regard to experimental observation on human ear. Moreover, the computational efficiency of our approach allows real-time interactions, making it suitable for use in a training simulator.

In middle ear surgery, the anatomy of the patient has a strong influence on the choice of surgical procedure or on a contraindication statement. To account for rehearsal simulation, information from the patient should be taken into consideration. A registration approach of our model to fit the anatomical data is developed and tested. The main advantage of this registration approach is that the process is performed on a tested and validated mechanical model. Indeed, all the mechanical parameters such as the meshing, the boundaries conditions, etc, are still implemented into our atlas after the registration. However, only the geometry of the atlas is adapted to patient data, the mechanical results provided by the registered models are similar to the atlas model.

To build a predictive patient specific model, we should also use the mechanical parameters (such as the Young moduli of each structures) that correspond to each individual patient. Indeed, these parameters differ from a patient to an other and the search of parameters from the patient data is still future work. Using such a predictive model, the simulation could find new applications. Namely, it could be used as a planning pre-operative tool to anticipate the choices that are currently done during the surgery: selection of ossicular bones to repair, type of implantable prosthesis, etc.

5 Conclusion

A mechanical atlas of the human middle ear was implemented for surgical training and rehearsal simulator. The realism of the atlas behavior was evaluated in physiological and in surgery condition. The computation efficiency of the developed atlas allows for real time computation simulation and for haptic interactions. To our best knowledge, this is the first paper that presents such a real-time simulation. An interactive method is reported to adapt the proposed mechanical atlas to the individual anatomy of the patient in few minutes. A patient specific model is obtained directly from the evaluated mechanical model avoiding the time-consuming work of a manual segmentation, mechanical parametrization and evaluation. The presented study is the basis of a microsurgical simulator design for training, practice and planning for otological surgery.

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