3D BIOMECHANICAL TONGUE MODELING TO STUDY SPEECH PRODUCTION

Jean-Michel Gerard[12], Pascal Perrier[1], Yohan Payan[2]

[1]Institut de la Communication Parlée, UMR CNRS 5009
Institut National Polytechnique de Grenoble & Université Stendhal, Grenoble, France

[2] Techniques de l'Imagerie, de la Modélisation et de la Cognition, UMR CNRS 5525
Université Joseph Fourier, La Tronche, France

ABSTRACT

The study of speech motor control from kinematic and acoustic signals collected from speakers requires an evaluation of the influence of the physical properties of the speech apparatus on the time variations of these signals. For this purpose, a 3D biomechanical tongue model is currently being developed at ICP. It is based on a 3D Finite Element mesh, in which 10 different tongue muscles are represented. The non-linear elastical properties of human tongue tissues are accounted for with hyperelastic characteristics and a large deformation modeling framework was chosen to accurately describe non-linear geometrical changes. The model is embedded in a realistic 3D geometrical description of the jaw and of the vocal tract walls, which are both considered as fixed and non deformable at the current stage of the model's development. In this paper, each modeling step is described and the impacts of the main tongue muscles on tongue shapes and vocal tract geometry are assessed through a number of various simulations. Their potential influences on speech motor control are discussed in the final section.

1. INTRODUCTION

Understanding speech motor control requires the collection and the analysis of numerous different kinds of articulatory and acoustic data from speakers. Unfortunately, experimental data cannot be the only basis for advanced our understanding of speech motor control theories for two main reasons. First, in spite of the fact that measurement devices and data processing tools have benefited from great improvements in the recent years, experimental data analysis still has its limitations. A simple illustration of this statement can be found in the analysis of EMG signals collected in parts of the tongue where muscle fibres are highly interwoven. Under such conditions, it is quite impossible to be sure about which muscle the signal comes from (for example, in the middle of the tongue where Transversalis fibres and Posterior Genioglossus fibres are interwoven). Second, in the measured signals it is often impossible to separate what is due to the Central Nervous System and what is due to the physics of the speech apparatus. As concerns for example EMG signals, the presence of mechanoreceptors in facial and tongue muscles contributing to muscle activation, makes it impossible to know which part of the EMG activity is due to descendent inputs from the Central Nervous System, and which part is the result of afferent inputs. From this perspective, biomechanical models are very useful, since they describe in detail muscle anatomy, muscle mechanics, muscle force generation mechanisms and the relations between muscle activation and tongue deformation. Thus, comparing simulations obtained with these models with experimental data provides a powerful approach to interpret the physical signals of speech production.

To our knowledge, the first biomechanical model used to study the impact of the physical properties of the articulatory system on speech production was designed by Perkell (1969). Perkell's model described the tongue in the mid-sagittal plane, and it was based on a discrete mass-spring modeling, including a distributed force model and tissue incompressibility. It also accounted for dynamics, and differentiated passive and active tissues. Simulations run with this model provided interesting insights into the relations between muscle activation and tongue deformation as well as into the constraints due to tongue and vocal tract morphology. Kiritani *et al.* (1976) achieved major improvements in developing the first 3D biomechanical model based on continuum mechanics and on Finite Element Method (FEM). This model also included a simplified lip model and was accounting for contacts between tongue and palate. Thus, the authors were able to study how tongue muscles are involved in the production of American English vowels.

The first model of the jaw and hyoid bone set was designed by Laboissière *et al.* (1996). They tested the influence of the jaw biomechanics on jaw displacements when constant motor command shifts were provided for different jaw positions in the mid-sagittal plane. They observed that the influence of the biomechanics was small in this specific case, and they concluded that there should be no need for the Central Nervous System to know precisely jaw biomechanics to control precisely this articulator.

Payan & Perrier (1997) designed the first 2D continuous biomechanical tongue model that includes elastical non linearities, the stiffness of the tissues varying with muscle recruitment, which both contribute to a more realistic description of tongue behaviour. The model is still extensively used to assess different potential impacts of tongue biomechanics on tongue movements and tongue shapes (Perrier *et al.*, 2003). Sanguineti *et al.* (1998) used a very similar approach to design a 2D model including the jaw, the hyoid bone and the tongue, with which they studied the muscles synergies associated with the production of movements along the main directions of deformation of the tongue. Dang & Honda (2004) designed an extended 2D model of the tongue and the jaw, representing these articulators in the mid-sagittal plane with a 2 cm thickness. This represents an intermediate step between 2D and 3D descriptions.

The most ambitious model ever built was by Wilhelms-Tricarico (1995). It is a 3D biomechanical model, featuring mechanical and geometrical non linearities, with both a mechanical modeling accounting for anisotropy and for interwoven fibers, and with a muscle activation model. Unfortunately, due to the conceptual complexity of the model, computational issues induced strong limitations of the use of the model in studying speech production. In spite of that a large number of modeling principles developed by Wilhelms-Tricarico (1995) served as the basis for the development of the model that will be presented in this paper.

2. MODELING APPROACH

2.1. A biomechanical model? Why

As discussed above, we conceive articulatory models as powerful tools to understand the control underlying the production of speech and to test hypotheses about the way physiological, articulatory and acoustical signals of speech are likely to be influenced by the motor commands sent by the Central Nervous System, on the one hand, and by the physical characteristics of the articulators, on the other hand. From this perspective, the central question is: which level of description of the physical reality is required for us to reach our objectives?

Articulatory models can be classified in two ways: geometrical and biomechanical. Geometrical models provide an accurate description of the relation between vocal tract geometry and positions of the articulators. The techniques underlying the design of these models are well-established and reasonably simple, and this makes them very appealing. However, geometrical models cannot describe the impact of motor commands on the time course of articulatory displacements, because they do not account for the causes and the constraints of these displacements, which are typically associated with force generation principles and mechanical properties of the articulators. These phenomena are intrinsically taken into account in biomechanical models. For that reason, biomechanical models are required. However, as exemplified by Wilhelms-Tricarico's work (1995, 2000), building realistic biomechanical models requires the use of complex mathematical techniques, which induce computational complexity with potential convergence problems and long simulation durations. The approach that was adopted during the design of the 3D biomechanical tongue model presented in this paper was guided by a constant search for compromise between accuracy of the description and computational complexity.

2.2. A continuous modeling approach

In biomechanical modeling, two main approaches are possible: one of them, called the "discrete approach", is based on the use of functional discrete models while the other, the "continuous approach", uses the laws of continuum mechanics to establish the equations of movement. The latter approach is generally favoured because it provides greater accuracy. The continuous approach is typically based on the Finite Element Method (FEM henceforth) (Zienkievicz & Taylor, 1994; Bathe 1982), even though it can induce an important computational load. In this method, the mechanical continuum is divided into elements defined by nodes, and the Lagrangian equations of movement are solved at each node. Then, thanks to interpolation functions between nodes, a solution can be computed in terms of deformation, displacement or stress, at any point of the continuum.

2.3. A 3D model

Tongue muscles are used to move and shape precisely the tongue, both in the mid-sagittal plane (determining in particular the position of the constriction) and in the coronal plane (controlling in particular the shape of the constriction). In addition, during speech production, the tongue is often in contact with palate or teeth. The contacts arising between tongue and palate in or close to the midsagittal plane were accounted for in previous 2D models, and this approach has allowed some aspects of stop production (e.g. Fuchs *et al.*, 2001) to be studied. However, lateral contacts could obviously not be accounted for with this kind of model. This is a drawback, since lateral contacts do influence tongue deformations in two ways: (1) there is a direct mechanical effect due to the introduction of additional friction or obstacles; (2) there is an indirect effect due to sensori-motor afference which can modify the motor commands. All those reasons led us to design a 3D model.

2.4. A dynamical model

Time variations of articulatory and acoustics signals are perceptually very relevant. We think that these temporal variations are the result of a combination of central and peripheral influences. Indeed, timing is an intrinsic component of the dynamics of a system due to stiffness, damping and inertia, while motor planning can specify the timing of the commands (extrinsic component). A model of speech production has to account for both aspects. As a consequence, our model is a dynamical model.

3. MESH DESIGN

3.1. Design of a tongue generic mesh

The mesh design process is completely described in Wilhelms-Tricarico (2000) and Gerard *et al* (2003). The key points are as follows.. A generic mesh was designed from the Visible Human Project^{TM} data set (http://www.nlm.nih.gov/research/visible/visible_human.html). It specifies the shape of the elements (Fig 1) taking into account the main properties of muscle anatomy such as the principal orientations of muscles fibres. Then, a more precise implementation of muscles was done in the mesh thanks to accurate anatomical data from Netter (1989), Miyawaki (1974) and Takemoto (2001) (see Gerard *et al.* (2003) for a comprehensive description).

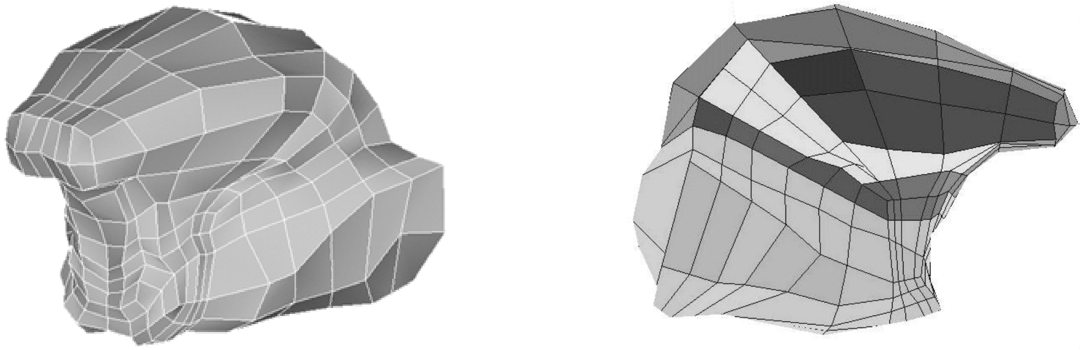

Figure 1: Generic finite element structure; left panel: generic mesh; right panel: muscular structures within the generic mesh; each level of gray represents a specific muscle; when several muscles are present in an element, only one colour is shown.

3.2. Adapting the model to a real speaker's geometry (mesh matching)

Evaluating a model of speech production implies the comparison of simulated movements, generated by the model, with articulatory data collected on speakers. At ICP, a large amount of various kinds of data were collected for a reference speaker (PB). In order to use those data, the geometry of the tongue model was adapted to PB's geometry.

The first step consisted of measuring carefully PB's 3D tongue contours at rest position were carefully measured from MRI data, with manual image segmentation. Then, a mesh-matching algorithm transformed the external contours of the generic mesh into a new mesh (henceforth called target mesh) reproducing PB' characteristics. Finally, the internal structure of the mesh was interpolated with respect to the transformed external contours (See Figure 2).

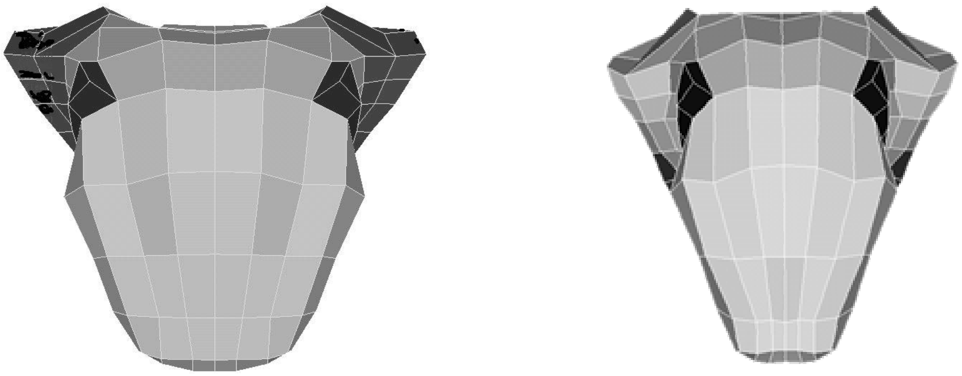

Figure 2: Top view of the tongue mesh before matching (left panel) and after matching with PB's tongue geometry (right panel)

3.3. Bony structures and vocal tract geometry

Since tongue positioning during speech production is made with reference to the vocal tract walls, a mesh of the whole vocal tract had to be designed. Once the tongue was matched to the reference speaker's geometry, meshes of the other structures such as jaw, soft and hard palate and hyoid bone were built. Several techniques were used. The jaw and the hyoid bone were manually extracted from CT-scan images, in the coronal plane. Then, spline functions were defined on each segmented image in order to obtain continuous representations of the contours. The software stacks up all these contours, creates a surface representation, and provides a mesh using the Marching cube algorithm (Lorensen & Cline, 1987). At this stage, the mesh describing the jaw contained approximately 15000 surface elements, which would generate, in case of contact between tongue and palate, a much too heavy computation load with unacceptable computation times. The ANSYS^{TM} FEM package (ANSYS, Inc., Canonsburg, Pennsylvania, USA, http://www.ansys.com/) was used to undersample the mesh with the constraint to carefully respect geometry in some crucial regions of the jaw, such as the region in the anterior part where the tongue is inserted in the jaw. The number of elements was reduced to 1450. The same technique was used for decimating the hyoid bone mesh.

The hard palate was designed from a dental cast which was digitized using a 3D scan. After undersampling, a 600 elements mesh was generated. The soft palate mesh was based on MRI data. A grid, made of regularly spaced planes, was used to spatially sample the vocal tract shape of PB at rest position. On each of those planes, the MRI data were interpolated and the soft palate contour was extracted. Then, using spline interpolation, a surface was created and a mesh was automatically set up.

The last step consisted of assembling all the components to make a unique mesh associating volume and surface elements. Note again that at the current stage of the modeling only the tongue is likely to move and to be deformed.

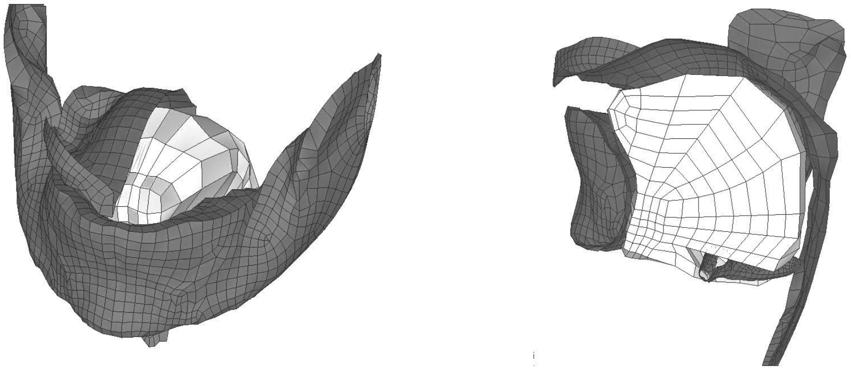

Figure 3: Complete 3D structure including tongue, hard palate, soft palate, jaw and hyoid bone in an oblique view (left panel) and in a mid-sagittal view (right panel)

4. MECHANICAL MODELING

On the basis of published force measurements during strong and sustained contacts between tongue and palate (Bunton & Weismer, 1994), and in the absence of any other kind of data, we estimated the muscle force level during speech production to be in the order of a few Newtons. With this level of force, the tongue displacement associated with a two phoneme sequence has to be as short as 50 to 100 ms. To reproduce this order of magnitude of forces and durations with the tongue model, an accurate estimation of mechanical properties of the tongue would be very helpful. Unfortunately, there's a lack of data in the literature about these properties. In order to increase the reliability of the model, we designed a specific indentation experiment to measure tongue tissues elasticity (Gerard *et al*, in press).

4.1. Indentation experiment

The goal of an indentation experiment is to obtain the constitutive law of a material, i.e. the relation between stress and strain, which determines the mechanical behaviour of the material. For this purpose, a human tongue was removed from the fresh cadaver of a 74 years old woman by Professor Lebeau from Grenoble University Hospital. The indentation experiment was run less than 24 hours after death, in order to limit tissue deterioration. It was done at the Purpan Hospital in Toulouse by Vincent Luboz from TIMC laboratory.

Human soft tissues, including those of the tongue, are likely to feature non linear elasticity, anisotropy, plasticity and visco-elasticity. As a first approximation, it was decided to model only non-linear elasticity, while visco-elasticity was considered to correspond to critical damping, and no account was given of anisotropy and of plasticity. Moreover, the tongue is covered with a layer of mucosa on its surface. In order to differentiate the mechanical properties of the mucosa from those of the tongue body, two indentations were successively made: the first was applied to the whole tongue (including the mucosa); for the second, the mucosa layer was removed and the remaining body was indented. In addition, the indentation was made at the rear and at the front of the tongue in each case, in order to test the possible limits of the isotropy hypothesis, primarily used in our model for computational simplicity.

4.2. Indentation results

The computation of a constitutive law from indentation measurements cannot be derived directly from the measurements. It was necessary to formulate hypotheses about the nature of the constitutive law and then to calculate the parameters of this law that permitted a good fit of the indentation measurements.

According to Fung (1993), hyperelastic material seems to be well adapted for a good approximation of human soft tissues. A material is said to be hyperelastic, if it is possible to find a function W, called the hyperelastic potential energy function (henceforth HPEF), whose derivative with respect to the strain equals the stress. The specification of this function determines the non-linear elastic properties of the material. Among the various strain-energy functions which can describe such a mechanical response (Holzapfel, 2001), we focused on the incompressible two parameter Yeoh strain-energy function W. This strain-energy function is given by the following analytical expression:

$$W = a_{10}(I_1 - 3) + a_{20}(I_1 - 3)^2 \quad \ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots\ldots (1).$$

where a_{10} and a_{20} are two constants that characterize tissue properties, while I1 is the first invariant of the right Cauchy-Green strain tensor C (I1=Trace(C)). Our objective was to infer from the indentation measurements a_{10} and a_{20} values which are compatible with the deformations measured on the cadaver tongue.

This was done by an iterative algorithm based on an analysis-by-synthesis technique. In this approach, a numerical model of the indentation experiment was designed (Figure 4), in which tongue tissues are modeled with a FEM structure and the indentor is simulated by the application of an external force. Starting from arbitrary initial values of the parameters determining the HPEF of the tissues, the iterative algorithm consists of (1) simulating the tissue deformations due to a given indentor force, (2) comparing the deformations with the experimental data, (3) updating the HPEF parameters and starting again with step (1), until a convergence criterion between simulated and experimental data is reached. This algorithm is described in more details in Gerard *et al.* (in press). Figure 5 shows an example of the results obtained with the optimization procedure: for the rear part of the tongue and without mucosa, the dotted line depicts the measured relation between force and displacement and the solid line corresponds to the simulated relation obtained with the optimal HPEF. It can be seen that the approximation is very accurate, since the error between the measured curve and the simulated one is smaller than the variance of the measures.

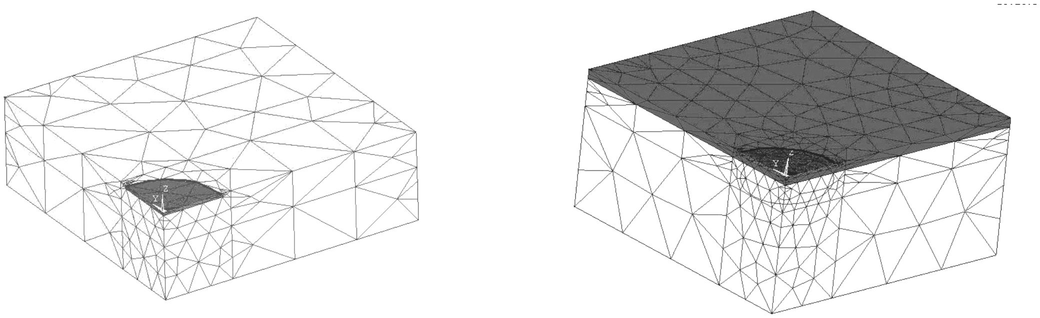

Figure 4: Meshes used for numerical simulations of indentation. Left: one layer model (tissues without mucosa). Right: two layers model (tissues and mucosa). The quarter circle represents the area where the indentor applies pressure.

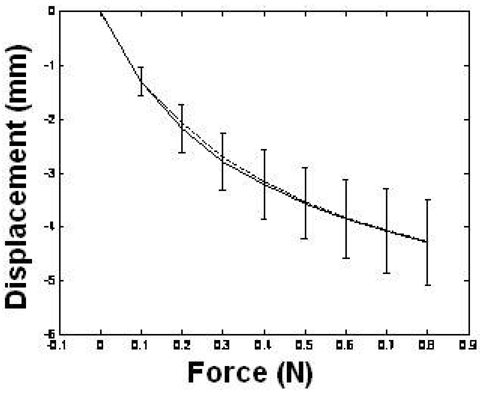

Figure 5: Measured (dotted line) and simulated (solid line) curves between indentor displacement and force. The standard deviation of the measure is depicted by vertical segments around the measured curve.

Table 1 gives the Young modulus which is computed as the derivative of stress with respect to the strain and corresponds to the stiffness of the material in the direction of the stress, inferred at rest position in the rear and front parts of the tongue, with and without the mucosa layer. It can be seen that (1) the rear and front values are very close, and (2) the mucosa does not significantly affect the global elasticity. These observations support our isotropic hypothesis.

Experiment	E
Rear muscular without mucosa	1.16 kPa
Front muscular tissues without mucosa	1.10 kPa
Rear mucosa	1.19 kPa
Front mucosa	1.14 kPa

Table 1: E is the inferred Young Modulus values for the cadaver tongue under the 4 different experimental conditions

5. SIMULATIONS

The methodology described above has allowed us to build up a tongue model whose mechanical properties are realistic. This was an important challenge of the current study, since our project is to use the model to infer the motor commands underlying tongue movements in speech. In this purpose, the model has absolutely to be reliable in terms of description of the physics. Since the model is adapted to speaker PB, future works in this project will consist (1) of a comparison between predictions made with the model and data collected from speaker PB for a number of sounds whose muscle activations are well-known, and (2) of attempts to infer muscle commands from tongue shapes measured during speech on speaker PB.

Before that it is necessary to determine the best approach for the solving of the Lagrangian equations of movement.

5.1. Small versus large deformation framework

The step by step solving of the Lagrangian equations of movement can be made within two different theoretical frameworks, which determine how geometrical deformations and strain are related to each other. If ∇D is the deformation gradient, the strain tensor is given by:

$$\varepsilon = \frac{1}{2}\left(\nabla D + (\nabla D)^T + \nabla D^T(\nabla D)\right) \tag{2}$$

If we assume that the system has small deformations, the second order term can be neglected. In this case, the strain tensor can be simplified, and it becomes a linear function of the deformation gradient. This corresponds to the so-called “small deformation framework”.

However, in some cases, the modeled continuum distorts sharply, implying higher values in the deformation gradient, and the above simplification is no longer valid. In this case, significant changes occur in the directions of loads during the movement and in stiffness values. Consequently, the stiffness matrix K and the load vector F depend on the displacement vector u. Hence, the equation to be solved for a static problem, is:

$$K(u) \cdot u = F(u) \tag{3}$$

This is the so-called “large deformation framework”. Iterative algorithms must be used to solve the equations.

Napadow *et al.* (1999) measured tongue deformation rates during speech production and they found values as large as 160 per cent in compression and 200 per cent in stretch. Since modeling deformations beyond 20 per cent are assumed to be large in soft tissues, a large deformation framework was required for our simulations.

5.2. Simulation settings

The tongue is assumed to be incompressible and to be critically damped. Damping was accounted for in the model using the Rayleigh damping method, where the damping matrix [C] is a linear combination of the mass matrix [M] and the stiffness matrix [K]:

$$[C] = \alpha[M] + \beta[K] \tag{4}$$

where α and β are the Rayleigh damping coefficients, computed to reach critical damping. Contacts are modeled for the hard palate and for the interior part of the jaw according to a penalty method. This method allows a small interpenetration of the materials in contact, and models the contact reaction force with a damped non-linear spring system. Contact parameters, such as contact stiffness and maximal allowed penetration were set in an ad-hoc manner, in order to obtain an acceptable maximal penetration (0.1mm) and to avoid numerical instabilities when contact occurs.

In the current version of the model, the jaw is fixed, and displacements of the hyoid bone are limited to the horizontal plane. Muscle forces are applied as external forces on nodes using a distributed force model (Payan & Perrier, 1997). Lagrangian equations of movements are solved using the FEM package ANSYS^{TM} with a combination of Newmark method and Newton-Raphson method (Gerard *et al.*, 2003). All simulations presented in this paper were obtained with muscle forces applied as step functions during 120 ms.

5.3. Results

Before precise and quantitative comparisons between simulations and data collected from speaker PB can be made, the model behavior was evaluated by observing the qualitative impact of specific muscle recruitments (Posterior Genioglossus, Anterior Genioglossus, Hyoglossus and Stylogossus) on the tongue shape. This is the purpose of this section. The results were qualitatively assessed by visually comparing the resulting tongue shape in the model with vocal tract X-ray images in the mid-sagittal plane (for French Bothorel *et al.*, 1986, and for English Perkell, 1969). To do so we selected among the X-Ray images those corresponding to sounds for which EMG data (e.g. Baer *et al.*, 1988) have suggested that they are mainly produced with the recruitment of one of these muscles.

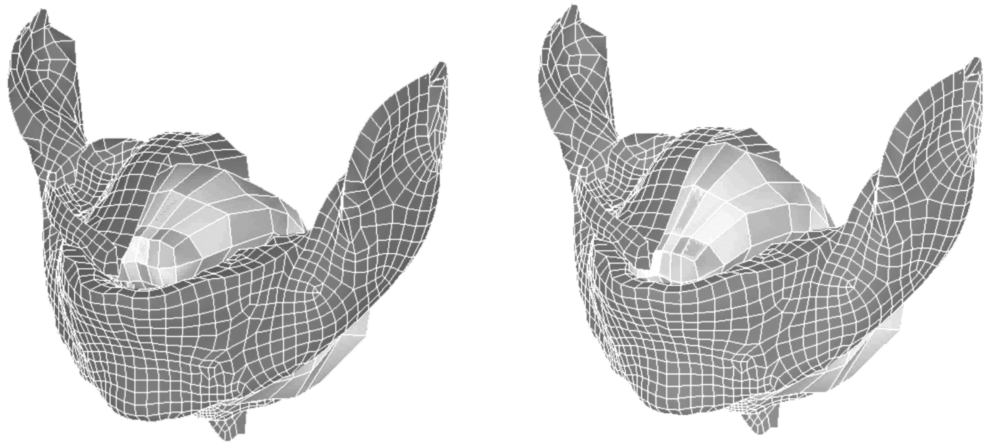

Figure 6: 3D tongue deformation (oblique front view) simulated on the model for a 1N force produced by the GGp Left panel: tongue at rest position. Right panel: Final tongue position. The light gray mesh represents the tongue. The dark meshes represent respectively the jaw (in front of the tongue and on its sides) and the right half of the hard palate (upper mesh).

Figure 6 shows the results obtained for a 2N force generated by the Posterior Genioglossus Posterior (GGp), which is known for its role in the production of high-front vowels such as /i/, /🖑/, and /e/. Our simulation shows a main rear-front movement of the tongue body with an elevation of the tongue dorsum due to contact with the incisors and due to tissue incompressibility. The simulated forward displacement of the rear of the tongue was about 2cm and the elevation of the tongue dorsum 0.9mm. This is consistent with X-ray data on vowel /i/ in English and French.

Changes in the coronal plane were also observed due to the fact that tongue tissues are assumed to be isotropic and incompressible. These latter changes can be limited by activating the Transversalis, since its fibres are oriented in the coronal plane. Combining an activation of GGp (2N) with an activation of the Transversalis (0.1N), it was possible to limit tongue expansion in the coronal plane which in turn increases the elevation of tongue dorsum from 9mm up to 1.3cm. This example shows how muscles are likely to influence the tongue shape in a plane which is orthogonal to the main orientation of its muscle fibers.

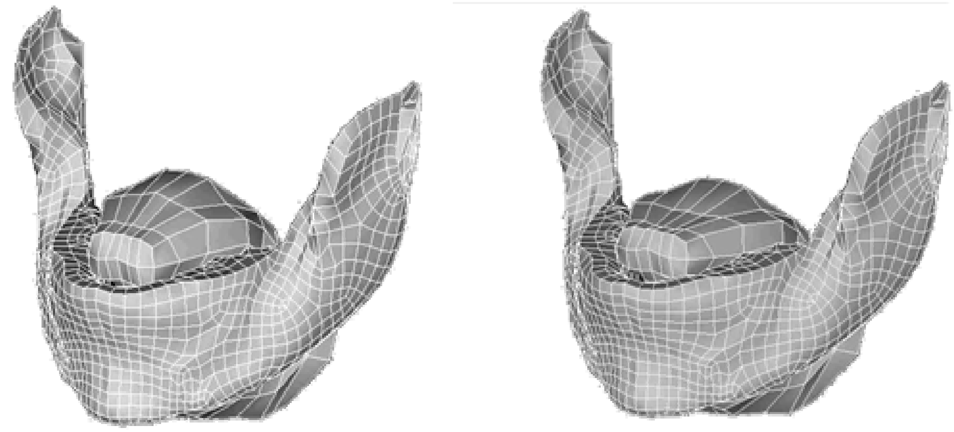

Figure 7: 3D tongue deformation (oblique front view) simulated on the model for a 0.5N force produced by GGa. Left panel: tongue at rest position. Right panel : Final tongue position. The mesh in front of the tongue and on its sides represents the jaw.

Fig 7 shows the tongue deformation generated by a 0.5N force produced by the Anterior Genioglossus (GGa). As expected (Baer *et al*, 1988), this muscle lowers the tongue tip slightly. This is in agreement with X-ray data on vowel /a/ that is known to be mainly produced with the combination of the GGa and the hyoglossus. It should also be noted that, in addition, it produces a slight tongue grooving in the apical region. This grooving is similar to the one observed on speakers during the production of /i/. It suggests that a combination of the recruitment of GGa, GGp and the Transversalis could underlie the production of /i/.

The consequence of an activation of the Hyoglossus in the tongue model is shown in Figure 8. Located at the rear of the tongue, the Hyoglossus lowers the back part of the tongue body and moves it back. As said above, this muscle is known for being involved in the production of the vowel /a/. The simulation was obtained for a 1N force. The vertical lowering of the rear of the tongue was 1.2cm.

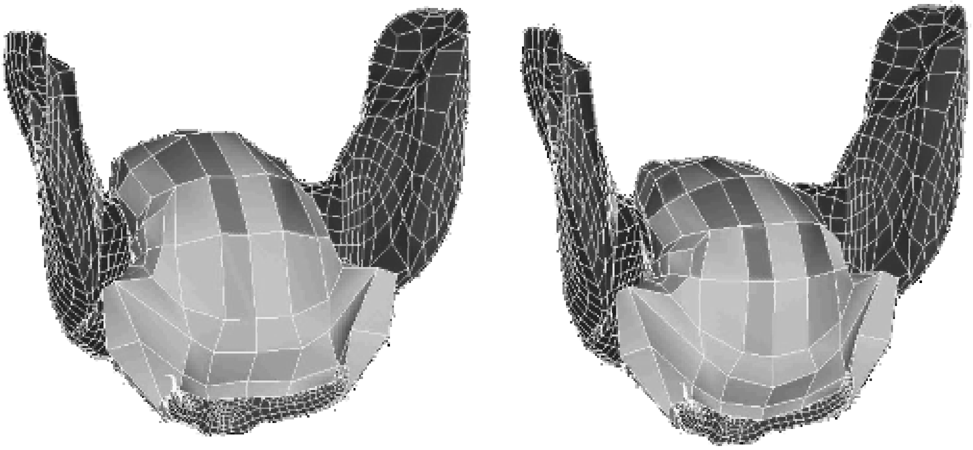

Figure 8: 3D tongue deformation (oblique view from the back) simulated on the model for a 2N force produced by the Hyoglossus. Left panel: tongue at rest position. Right panel : Final tongue position. The light gray mesh represents the tongue. The mesh in front of the tongue and on its sides represents the jaw.

Figure 9 shows the deformation associated with a 1N force produced by the Styloglossus. An elevation of the tongue in its dorsal region can be observed together with a lowering of the tongue tip and a general backward displacement of the whole tongue body. This is very close to the tongue shape observed for velar vowels such as /u/. The direction of the deformation is consistent with EMG data (Baer *et al.*, 1998) about the role of the Styloglossus in the production of velars. However, sharp edges can be observed on the top of the tongue (right panel), which is not realistic. This suggests that the muscle force could be applied too locally on specifc elements of the tongue, namely the ones where the external fibers of the styloglossus are inserted. This has to be further assessed.

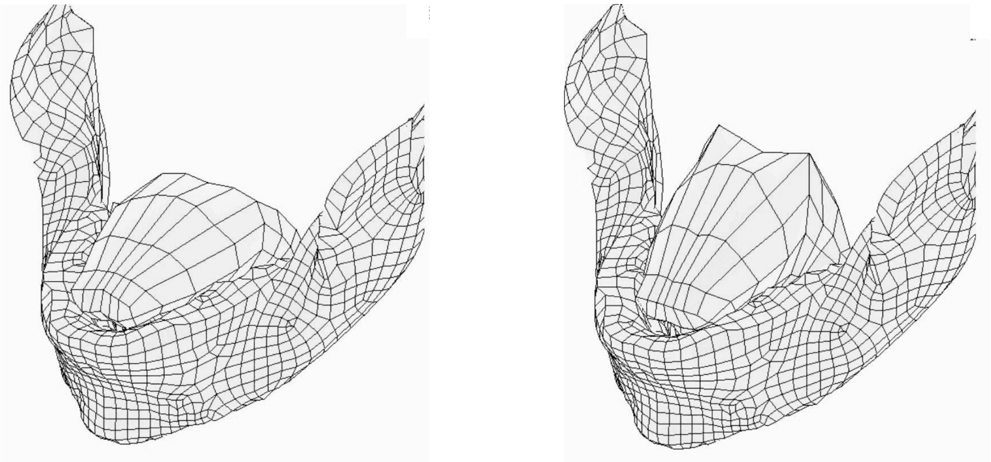

Figure 9: 3D tongue deformation (oblique front view) simulated on the model for a 0.5N force produced by the Styloglossus. Left panel: tongue at rest position. Right panel : Final tongue position. The mesh in front of the tongue and on its sides represents the jaw.

6. CONCLUSION

A 3D tongue model was developed to study speech production and to test motor control models in speech production. Its design takes into account some complex aspects of the mechanical characteristics of tongue tissues, such as non-linear elasticity and large deformations. An experimental indentation allows a realistic estimation of the elasticity parameters. Thus, tongue shape variations compatible with observations on human subjects during speech production have been simulated with reasonable muscle force levels and with realistic movement durations (less than 100 ms).

In future analyses, the model, representing a real speaker's geometry, will be validated using MRI data for French phonemes and the related acoustic output from the same speaker.

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