Simulation of 3D larynges with asymmetric distribution of viscoelastic properties in their vocal folds

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Abstract

Here 3D larynx models whose viscoelastic properties of the superficial tissues of the true vocal folds were altered in specific regions at the glottis - producing different left-right fold asymmetries - are analyzed. The idea is to initiate studies about the dynamic of larynges under possible pathological conditions. All simulations used a strategy based on the finite element method to solve a fluid-structure interaction (airflow in the larynx) with a contact-impact problem (glottal closure). The results show that glottal closure is directly influenced by the presence and location of a stiffer mass in the surface of one of the vocal folds. Changes in glottal signal F_0 (up to $+13.7$Hz), open quotient (up to 0.5812), and relative jitter (up to 15.2381%) were observed and are discussed here.

Index Terms: finite element method, larynx, simulation

1. Introduction

Self-sustained oscillations in fluid-structure interactions take place when specific physical conditions are reached. In the case of human phonation, a periodic movement of the vocal folds (VFs) depends on the glottal gap area, the translaryngeal pressure drop, the viscoelastic properties of the tissues in both VFs, and other conditions. Moreover they define if the resulting glottal signal will have enough harmonic content (due to the ability of the VFs to close the glottis) to excite the supra-glottal cavity and to produce voice signals [1, 2, 3].

Laryngeal diseases such as unilateral VF paralysis, polyps, and Reinke's edema, for example, change the VF tissue properties and lead to incomplete glottal closure or abnormal movements of true VFs due to the change in the physical conditions that self-sustain a periodic VF oscillation. By modeling the larynx under these conditions [4, 5, 6] or by analyzing the voice signal to extract some numeric parameter related to any physiological disturbance [7], such diseases have been objectively analyzed.

Previous works [5, 6, 8, 9] used simulations to compute the effects of asymmetric VFs over the glottal signal, which are related to VF paralysis. Using a 2D-immersed boundary method to deal with the moving boundaries of the airflow, a finite difference method to solve unsteady, incompressible Navier-Stokes equations, and an adapted two-mass model to represent an asymmetric larynx, [9] related F_0 and phonation offset to differences in their tissue representation of both VFs. Recently asymmetries [5, 6] were added into nonlinear equations that described the surface motion of the VFs in order to evaluate their effect into vocal jitter.

The present work numerically investigated the behavior of larynges whose superficial tissues (Hirano's cover [10]) of one of their VFs have their viscoelastic properties altered in comparison to the rest of the larynx tissues, comparing it with similar strategies [6, 9, 11]. Physiologically these localized changes represented preliminary attempts to simulate laryngeal diseases (like sulcus vocalis, for example) with similar tissue modifications, and to verify their influence on the VFs movement and the volumetric velocities (glottal signal).

Such unbalanced larynges were numerically simulated through a conventional finite element method (FEM) procedure [12, 13, 14] which includes numerical descriptions of both the laryngeal airflow and the contact forces that appear in the glottal surface during collisions of both VFs. This mathematical approach has been used in other works [3, 9, 11, 15].

2. Material and Method

The simulation involved the interaction between two media: the tissue medium (whose viscoelastic description follows Hirano's body-cover model), and the airflow medium (described by Navier-Stokes equations [16]). The glottal closing phase was represented by a contact-impact model (similar to the ones used in computerized car-crash tests [12]). All these fluid-structure interactions were simulated using a FEM computer code developed by the author ([13]).

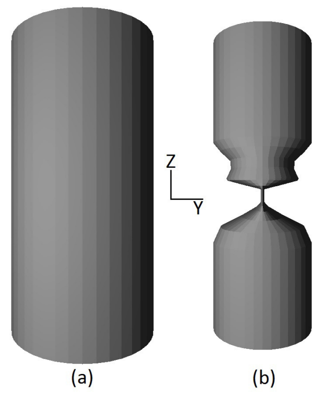

Figure 1: Bounding surfaces used to define both larynx tissue and airflow channel. The larynx tissue was limited by external (a) and internal (b) surfaces while the airflow medium was limited by the internal surface. Both cylindrical boundaries are 4-cm long.

2.1. Airflow and tissue meshes

The airflow and tissue meshes were constructed from two distinct surfaces: one separating both media, and another surrounding the larynx tissue. The latter one - a 3D cylinder - was used to rigidly sustain the simulated larynx (Figure 1). Both surfaces were described by 3D ellipses [13]. An automatic mesh generator generated all tetrahedrons used by FEM [12, 16, 17].

The larynges were defined from three different tissues, namely cover, ligament, and body [10]. A fourth tissue was introduced to unbalance the VFs. Geometrically it started at anterior commissure, ended at posterior commissure, and was positioned only in one of the VFs (called here as "modified" fold). It was also in contact with the airflow and, naturally, under direct influence of the airflow pressure. In Figures 2 and 3, "stiff_a" and "stiff_b" were two tissues that had these properties. They were positioned at different heights of the simulated larynx in order to verify the influence of their locations over the larynx dynamics (respectively at the end and at the center of the true VF). The exact geometric description of the simulated tissues is found in [13].

2.2. Tissue properties

The viscoelastic properties of the regular tissues (cover, ligament and body) were defined according to [1, 18, 19] who made extensive studies with canine and human larynx tissues. They are shown in Table 1.

Table 1: *Viscoelastic properties of the laryngeal tissues (E, E', and μ' are given in* $^{\mathrm{kdyn}}/_{\mathrm{cm}^2}$. *$\eta$ is given in* poise. *ν and ν' are dimensionless quantities).*

Tissue name	E	ν	E'	μ'	ν'	η
Cover	20	0.76	200	200	0.76	10
Ligament	40	0.68	200	400	0.68	10
Body	200	0.45	200	300	0.45	150
"Stiff_a" "Stiff_b"	100	0.56	200	300	0.56	150

The density (ρ) of all tissues were set to $1.03\ ^{\mathrm{g}}/_{\mathrm{cm}^3}$. Transverse Young's modulus (E) for cover and ligament tissues were respectively set to 5 and 10 times lower than for body tissue. Along with longitudinal Young's modulus (E') set to $200\ ^{\mathrm{kdyn}}/_{\mathrm{cm}^2}$, they contributed to keep the true VFs almost rigid under aerodynamic pressures produced during phonation. Transverse and longitudinal Poisson's ratios were defined in order to keep the tissues nearly incompressible. [20, 21] allowed larger transverse movements of body tissue than the ones presented here. The values in Table 1 assumed that during phonation the true VFs were transversely strengthened in order to reduce the glottal area, allowing only that material waves traveled over the cover tissue. Of course there was a transient phase when the whole VF bodies moved upstream in order to reach a stationary position [13].

Viscoelastic properties of the tissues "stiff_a" and "stiff_b" were chosen to represent stiff materials (resulting from a superficial trauma, for instance) placed over the glottal surface. Therefore the traveling material waves would be blocked when they tried to pass through these tissues. They were also kept near incompressible [22].

2.3. Airflow properties

The viscosity (μ_{f}) and density (ρ_f) of the laryngeal airflow were set to $0.000179\ ^{\mathrm{dyn.s}}/_{\mathrm{cm}^2}$ and $0.00123\ ^{\mathrm{g}}/_{\mathrm{cm}^3}$, respectively. The translaryngeal pressure drop (between lung and mouth) was set to $+8000\ ^{\mathrm{dyn}}/_{\mathrm{cm}^2}$ and the airflow was assumed to be incompressible under phonation conditions [14].

2.4. Numerical procedures

Over time, the airflow volumetric and tissue displacement were calculated by the following steps:

1. Solve Navier-Stokes equations [16] for the airflow;
2. Gather air pressures at the air-tissue interface;
3. Calculate tissue displacements [1, 12];
4. Calculate collision forces (when they exist) [12];
5. Update tissue and airflow meshes [23].

The pressures and air velocities at the airflow nodes were obtained from solving Navier-Stokes equations (1) by FEM. In addition to having set a pressure drop to excite flow through the larynx model (Neumann boundary condition), the air velocities at the larynx walls was set to zero (Dirichlet boundary condition).

$$\begin{aligned} \nabla \cdot \vec{v} = 0 \\ \rho_f\ (\vec{v} \cdot \nabla)\vec{v} = -\nabla p + \mu_f\ \nabla^2 \vec{v} \end{aligned} \tag{1}$$

The tissue displacements were obtained by solving time-dependent mechanical equations [1, 13] by FEM (using Newmark method [13] because it is unconditionally stable, and a time step set to $300\mu s$ for all simulations). It was required setting the tissue displacements at all external surfaces of the larynx tissue mesh to zero (Dirichlet boundary condition), except at the surface of the mesh in contact to flow mesh. There the forces resulting from the calculation of airflow pressures defined the Neumann boundary condition to be used in tissue displacement calculation.

The material properties for all the simulated laryngeal tissues obeyed a linear stress-displacement relationship. The reasons for that were: 1) the difficulty of determining the true relationships of the viscoelastic material [24]; 2) reducing the complexity in simulating the larynx model. These materials were considered transverse isotropic [19, 22, 15] to capture the fibrous nature of the VF tissues.

The collision forces ($\vec{f}$) were obtained by calculating the amount of positive normal forces ($\lambda \geq 0$) needed to avoid that the VFs to penetrate each other. The procedure [12, 13, 17, 25] consisted of adding inequalities in the form of displacement restrictions (shown in 2) to all tissue nodes ($\vec{x}$) that penetrate a tissue surface at $\vec{y}$. Such restrictions should impeded body interpenetration ($\mathrm{gap}(\vec{x}) \geq 0$). This was an iterative procedure because it was not *a priori* known which tissue mesh nodes would collide with a tissue surface.

$$\begin{aligned} \mathrm{gap}(\vec{x}) = (\vec{x} - \vec{y})' \cdot \vec{n} \geq 0 \\ \lambda = (\vec{f} \cdot \vec{n}) \geq 0 \\ \mathrm{gap}(\vec{x})\lambda = 0 \end{aligned} \tag{2}$$

In Eq. 2, $\vec{n}$ is the vector that is normal to the penetrated tissue surface. This contact-impact problem demanded that both the inequalties in Eq. 2 be mutually exclusive: either a laryngeal gaps or the collision forces are positives [12, 25], which means $\mathrm{gap}(\vec{x})\lambda = 0$.

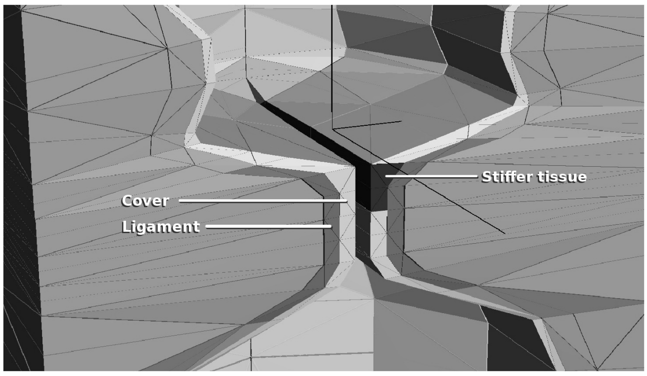

Figure 2: Cross section of the 3D larynx with tissue "stiff_a".

3. Results and Discussion

Two larynges were simulated respectively with tissues "stiff_a" and "stiff_b". Their placements in the larynges were chosen to verify their influence over the VFs movement and the glottal signal. Figures 2 and 3 illustrate such arrangements.

If the VF tissue is stiffer near the end of the glottis ("stiff_a" for example), it interrupts the propagation of the material wave traveling upstream in the VF surface [26]. Figure 4 shows a comparison between glottal signal collected from simulated larynges whose VFs had stiffer tissues, and the glottal signal from a normal simulated larynx [13]. For a larynx with tissue "stiff_a", the glottis was never closed; therefore the simulated larynx generated a glottal signal with reduced harmonic richness (resembling a sinusoidal wave).

There was no collision between both VFs because the material wave did not reach to the end of the VF (only the cover tissue of the normal VF had significant displacement). The amplitude of the material wave formed over the surface of the normal VF was significant but not enough to close the glottis once the amplitude of the material wave in the modified VF was small. It naturally reflected on a reduction of the glottal signal amplitude (Figure 4).

When a stiff tissue was placed downward the modified VF (Figure 3), the larynx dynamics allowed the glottis to be closed. The shape of the glottal signal was similar to the one obtained

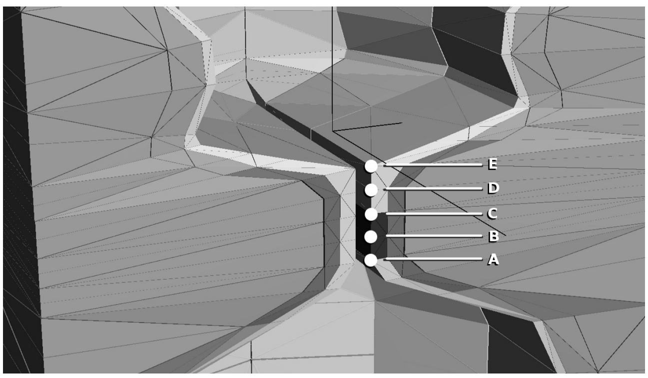

Figure 3: Cross section of the simulated larynx with tissue "stiff_b". A, B, C, D, and E are nodes (located over the modified fold, 0.07cm apart from each other) whose displacements were tracked.

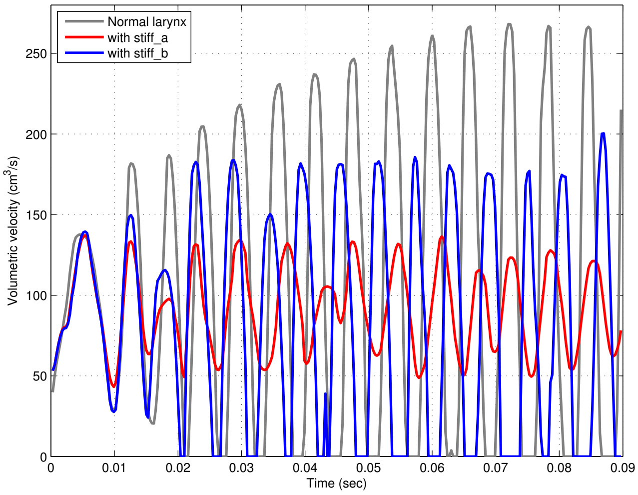

Figure 4: Comparison of glottal signals captured from simulations of larynges with tissues "stiff_a" and "stiff_b". The reference glottal signal was from a simulated normal larynx.

from a normal larynx, although its F_0 and amplitude have been respectively increased and reduced. The material wave over the modified VF started just after the stiff tissue, having a short spatial interval (0.14cm in rest position) to travel to the glottis end. For comparison the F_0 and open quotient for the normal larynx was 164Hz and 0.6247 while for the larynx with tissue "stiff_b" was 177.7Hz and 0.5812, respectively. It gave an increase of 13.7Hz by moving the stiffer tissue 0.14cm upward. These results are in typical range of larynx under similar conditions as reported by [11].

The influence of having the stiffer tissue at different positions in the surface of the modified fold is also shown in Figures 5, 6, 7 and 8. These positions were chosen to capture the material wave formed over these folds. At first, the presence of a stiffer tissue at the end of the modified fold significantly reduced the tissue displacement.

The phase difference between the nodal displacements of the nodes D and E (both of them located closer to the end of the VF, and 0.07cm apart from each other) was of $0.68\ ^{\text{rad}}/_{\text{sec}}$ for the modified fold with "stiff_b" tissue while there was not significant phase difference for the modified fold with "stiff_a". This happened because this tissue was stiffer than the regular cover tissue and did not deform enough. A comparison with two-mass model of the larynx [2], whose masses were 0.15cm apart from each other and produced a phase difference of $0.96\ ^{\text{rad}}/_{\text{sec}}$, revealed that the tissue at the end of the VFs should be soft enough to create a glottal material wave.

The lack of deformation of the stiffer tissue was observed in the displacements at nodes A, B, and C for tissue "stiff_b" and also in Figure 8. Their magnitudes were comparable to the one obtained for nodes D and E in "stiff_a" tissue. Surrounded by softer tissues, these stiffer masses acted as barriers to superficial movements of the modified fold.

The asymmetric behavior of the material waves in both VFs produced a glottal signal with irregular peak-to-peak amplitudes (mainly in simulations with tissue "stiff_a"). In the case of the larynx with tissue "stiff_b", the open-quotient was irregular in comparison to normal larynx (Figure 4). The glottal relative jitter for such a larynx presented a significant increase, reach-

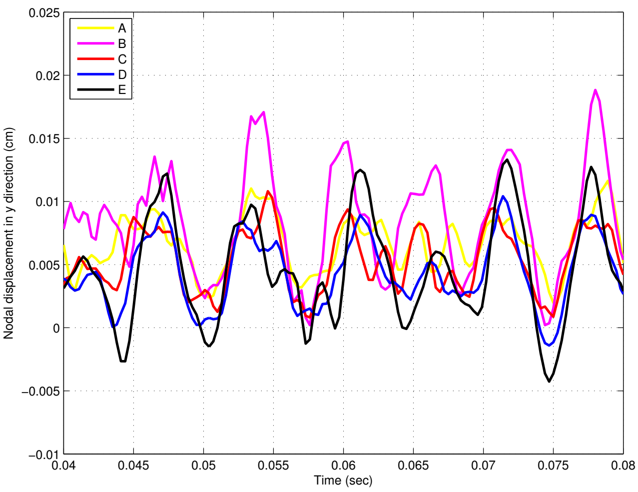

Figure 5: Displacement of nodes A, B, C, D, and E, in the larynx with tissue “stiff_a” (Nodal locations in are shown Figure 3).

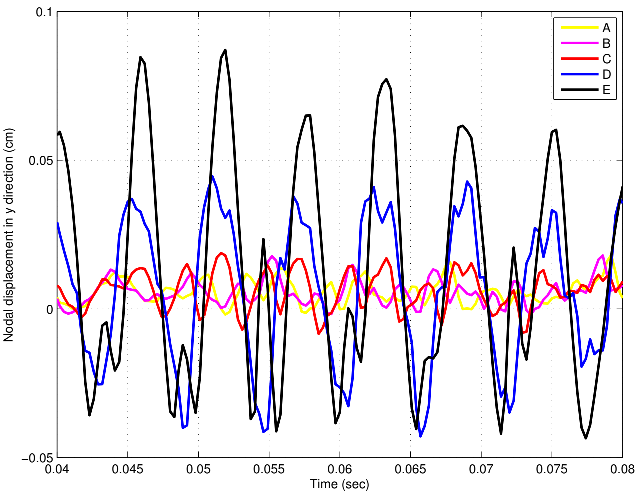

Figure 6: Displacement of nodes A, B, C, D, and E, in the larynx with tissue “stiff_b” (Nodal locations are shown in Figure 3).

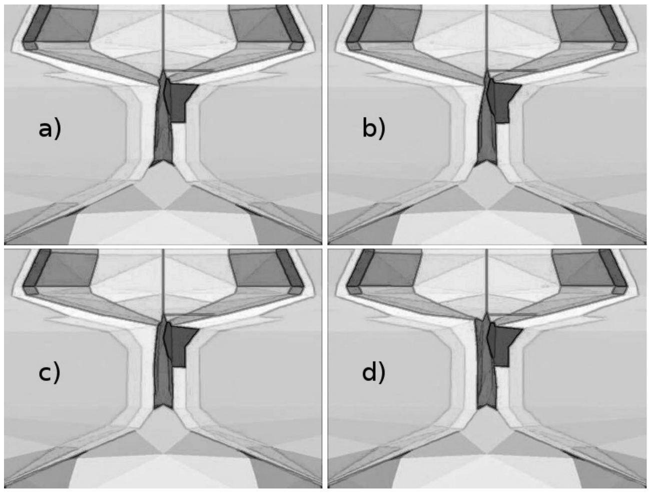

Figure 7: Snapshots of transverse view of the larynx with tissue “stiff_a” at a) 64.2, b) 65.1, c) 66.00, and d) 66.9ms.

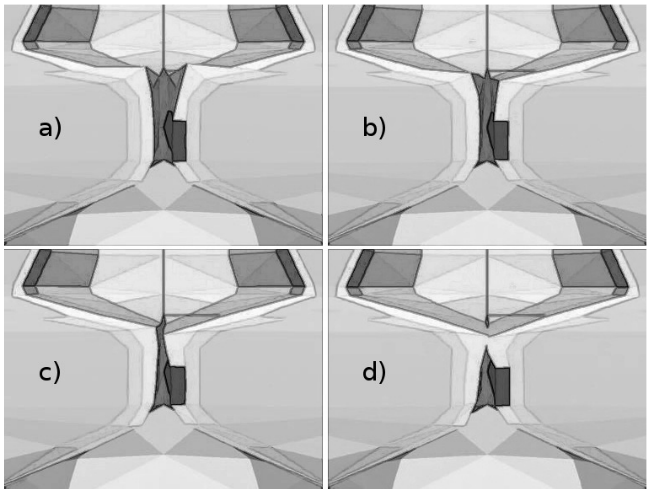

Figure 8: Snapshots of transverse view of the larynx with tissue “stiff_b” at a) 39.9, b) 41.4, c) 42.0, and d) 42.6ms.

ing 15.2381%, higher than the ones obtained with simplified models for larynges in similar experiments [6]. However their analysis only considered phase-locked nonlinear systems.

It was observed that the VF collision acted as a tissue movement regulator, avoiding formation of very chaotic superficial material waves near the glottis. As pointed out by [6, 11], simplified models may not account such complexity.

4. Conclusions

The current study suggests that changing the cover tissue composition in a larynx can impede the glottis closure and therefore reduce the harmonic richness of the glottal signal that would be modulated at the supra-glottal structure. It also affects F_0, the relative jitter, and the open quotient of the glottal signal depending on the spatial position of a stiffer tissue in the VF surface.

Laryngeal diseases (like nodules, polyps, and Reinke's edema) and some surgical procedures applied over the VF surface change the larynx tissue constitution. The location of such interferences in the glottal surface is crucial to determine the ability of the larynx to produce a quasi-normal glottal signal. As seen here, if the cover tissue becomes stiffer at the glottis exit, the material wave can be blocked and there will not be a complete glottal closure.

Asymmetric larynx (in terms of their tissue constitution) can generate a quasi-normal glottal signal if both VFs can reach each other during the phonation. The cost is the presence of some abnormal movements along the phonation cycles, irregular open quotient, and reduction of the glottal signal amplitude.

Further studies will be conducted to establish a precise relationship between the composition (and spatial position) of the stiff tissue and glottal signal and VF displacements.

5. Acknowledgements

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6. References

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