

Solving for unsteady airflow in a glottal model with immersed moving boundaries

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Received 18 October 2006; received in revised form 21 June 2007; accepted 25 June 2007

Available online 20 July 2007

Abstract

This work builds on previous efforts to characterize the dynamic development of the airflow in the glottis from a fluid mechanical point of view. A multigrid finite-difference method with immersed boundaries is implemented to solve the Navier–Stokes equations in a channel constricted by a vibrating rigid structure with a shape conforming to the human vocal folds. For the dynamically evolving boundaries we apply a forced oscillation glottal model. The large scale deformations of the boundaries are handled without regridding and tracheal input velocity is either set to a constant value or synchronized with wall motion. Particular attention is paid to the mobility of the point where the airflow detaches from the flapping walls. Results illustrate the relevance and the diversity of flow separation dynamics within the constriction standing for the glottis, while flow instabilities past the constriction are not found to affect flow behavior between the moving walls significantly. A comparison between static and dynamic numerical experiments show that mobility of the flow separation point is nontrivial in general and only rarely quasi-static.

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Keywords: Navier–Stokes equations; Glottal airflow; Moving boundaries; Vocal-fold model; Flow separation

1. Introduction

Voiced sounds in human speech are produced by the pulsatile flow generated during the self-sustained oscillations of the vocal folds. Several issues are involved in an accurate description of the physics of this phenomenon, namely, the mechanical description of vocal fold oscillation, the aerodynamics of the airflow and the acoustics of the generated sound.

The first models for the mechanics of vocal fold vibration can be traced to the pioneering work of Ishizaka and Flanagan [1]. Their famous two-mass model for the motion of the upper and lower parts of the vocal folds has been considerably improved, but most of its mechanical features are still shared by the new generation of lumped models, which use a smoother layout [2] and a variable number of masses [3] to support the nonuniform deformation of the glottal channel geometry. The picture of glottal airflow within most of these mechanical models is based on the stationary Bernoulli equation, modified to account for viscosity effects, air inertance or pressure recovery downstream of the glottal outlet.

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A thorough revision of the fluid mechanical description of the airflow through the glottis is proposed in 1994. In this work [4], Pelorson et al. show that the standard flow portrait adopted by most low-dimensional lumped models is inadequate if the main volume-control effect, which is flow separation near the glottal outlet combined with turbulent dissipation of kinetic energy in the resulting jet flow, is disregarded. While many effects, such as flow separation at the entrance of the glottis, flow reattachment at the glottal outlet, or asymmetry in the flow due to the Coanda effect, do not appear to be relevant for phonation, the moving flow-separation point in the diverging channel does affect the resulting glottal volume signals, both qualitatively and quantitatively. Recent implementations of two-mass vocal fold models (such as [2]) include some flow separation criterion to account for the changes introduced by this flow model. However, the boundary-layer theory necessary to predict flow detachment is substituted with a geometrical criterion that assumes that flow separation is mainly sensitive to the degree of divergence of the glottal channel (known as the Liljencrantz's condition). The acoustic correlate of these criteria in the model's output is assessed in [5].

The dynamics of pulsatile flow with flow separation, is certainly a very general problem in fluid mechanics. In most cases, it can only be solved using the numerical techniques developed in computational fluid dynamics (CFD). Glottal flow is not an exception. Since the mid-nineties, different approaches of varying complexity have been implemented to solve the unsteady laryngeal flow by means of the direct numerical simulation (DNS) of the Navier–Stokes equations. For static models of the vocal folds, recent studies show that the flow can be satisfactorily approximated by asymptotic models, such as the boundary layer theory or the reduced Navier–Stokes equations [6]. Nevertheless, wall movement, which is known to have a strong effect on the airflow dynamics, justifies the effort of working with the full Navier–Stokes equations. Both, experimental and numerical studies [4,7], indicate that the intraglottal pressure distribution is highly sensitive to wall motion. Since the pressure distribution on the glottal walls determines the driving force acting on the mechanical structure, such effects need to be properly weighed, in view of the development of less caricatural biophysical models.

In addition, CFD methods applied to glottal flow may be helpful to study the aeroacoustics of phonation. While flow in the glottal constriction is locally incompressible (typical Mach numbers for phonation are of order 10^{-1}), it is well known that nonacoustic fluid motion can provide a source of sound. In fact, the glottis behaves as an acoustically compact source (its size is small compared to the acoustic wavelength) which adds mass into the vocal tract in a periodic way. In simplified vocal-fold models, the vocal source is plainly considered as an inertive local wave emitter (the radiated sound pressure is often approximated by the derivative of the airflow volume velocity). However, the exact nature of flow-sound interaction is less well understood. Previous studies have postulated that, along with the monopole source associated with the pulsatile jet, a dipole source arising from the interaction between vortical structures and the velocity field should be acoustically relevant in voiced sound production [8], at least at low frequencies. Since sound pressure at low Mach numbers is not predicted with sufficient accuracy with most Navier–Stokes flow solvers, it is customary to use a two-step procedure in which, the unsteady incompressible flow solutions are used to compute the aerodynamic sources, used in the second stage to compute the propagation and radiation of sound [9,10]. These methods rely on some acoustic analogy, which assumes that sound generation and propagation are decoupled (e.g. Lighthill's analogy [11]). Simultaneously solving for both, the flow and the acoustics, from the compressible form of the Navier–Stokes equation is considered impractical in most problems, and in particular at low Mach numbers. An attempt has been anyhow made in [12,13], with an artificially increased Mach number, a manoeuvre that remains controversial for a fluid flow in which the order of magnitude of the acoustic fluctuations is much smaller than other variations in the fluid variables.

In this context, computation of full numerical solutions for the incompressible flow within the glottal channel stays an essential theoretical tool to improve our knowledge on the physics of the flow-induced oscillations of the vocal folds, along with its structural-acoustic response. The trend in the last years is to perform numerical simulations validated by experimental setups, which reproduce some aspects of vocal fold behavior [14–16,6,17]. Other studies have compared Navier–Stokes solutions with simplified theoretical models or with previous experimental data. Further experimental work calling for sound theoretical tools is also in progress [18]. While most numerical studies (including those performed with CFD commercial codes) use a boundary-conformal unstructured grid for the fluid flow in the glottal channel, a structured grid for a flow with immersed boundaries (so that the solid boundaries move freely through the mesh) presents significant advantages for the dynamic problem. A few studies with low spatial resolution [19] or with simplified wall movement (channel with a moving indentation) [20] have been performed with this technique.

In this work, we retrieve the application of a multigrid finite-difference method with immersed boundaries [21], in order to study some of the particular features of flow dynamics through a constriction animated with the characteristic

flapping motion realized by the vocal folds during the production of voiced sounds. The fixed Cartesian grid is spared from the task of adapting to the moving boundary, thus complying with the large scale deformations of the flow boundaries imposed by vocal-fold motion. As a result, the computational cost is lower, and in addition, there is guarantee that the discretization scheme will not be degraded by nonorthogonal frames.

In this work, we will mainly focus on the airflow dynamics induced by wall motion within the rigid walls of the constriction in the absence of fluid-structure interaction. A systematic analysis of sensitivity to the parameters of the numerical model or of other flow variables (such as intraglottal or wall pressure) will be presented in a separate work. Our scope in this paper is to show how the flow tube that is generated in the flapping constriction can present diverse dynamics in different scenarios. The structure oscillations are prescribed (and thus forced on the system) along with boundary conditions of the problem. Notice that even if real vocal folds collide when vibrating, complete closure of the boundaries is deliberately avoided in numerical experiments (strictly speaking, taking collision into account would imply working with two separate numerical domains). In fact, we expect to implement numerical tests in a colliding numerical model with this technique in the near future.

2. Numerical model

A two-dimensional CFD method, is proposed to simulate airflow within the vibrating channel.

2.1. Governing equations

Noting $V = (U, W)$, the dimensionless continuity and Navier–Stokes equations are written as:

$$\frac{\partial V}{\partial t} + (V \cdot \nabla)V = \frac{1}{Re} \nabla^2 V - \nabla P, \quad (1)$$

$$\nabla \cdot V = 0. \quad (2)$$

The Reynolds number Re in our numerical model is built on the channel's maximum height H and the incoming axial velocity U_0 .

$$Re = Re_H = \frac{U_0 H}{\nu}. \quad (3)$$

2.2. Boundary conditions

In the literature, the incoming (tracheal) velocity is sometimes synchronized with wall motion and sometimes kept constant, a choice which mimics the constant subglottal pressure exerted by the lungs during phonation. Our numerical code can support both choices through an input parameter which turns on and off the synchronization.

On the solid (rigid) walls, we impose the classical no-slip and no-injection condition. As in [21], at the downstream boundary, we impose an advective-type boundary condition:

$$\frac{\partial V(\Gamma, z, t)}{\partial t} + \frac{\partial V(\Gamma, z, t)}{\partial x} = 0 \quad (4)$$

which reduces to a homogeneous Neumann condition if the flow reaches a steady state. No symmetry condition is forced for the flow along the channel axis (we have chosen to perform our calculations on the whole domain to capture the flow asymmetries detected in preliminary tests and in view of future applications).

2.3. Numerical method

The equations are integrated in unsteady form. The time marching algorithm is based on a prediction projection method which consists in computing an intermediate velocity field V^* assuming a known pressure field.

$$\frac{3U^* - 4U^n + U^{n-1}}{2\delta t} + [2(V^n \cdot \nabla)U^n - (V^{n-1} \cdot \nabla)U^{n-1}] = -\frac{\partial P^n}{\partial x} + \frac{1}{Re} \nabla^2 U^* \quad (5)$$

with a similar equation for W^* for the z coordinate. This scheme yields to independent Helmholtz problems for each component U^* and W^* . The velocity field V^* is not yet set divergence free. Projection onto the space of divergence

free vector fields is performed in a second stage, through a scalar variable Φ which can be seen as a pressure correction, such that:

$$V^{n+1} - V^* = \delta t \nabla \Phi. \quad (6)$$

To compute Φ , we take the divergence of Eq. (1), so that, for $\nabla \cdot V^{n+1} = 0$:

$$\nabla^2 \Phi = -(\nabla \cdot V^*)/\delta t \quad (7)$$

associated with the boundary condition $\partial \Phi / \partial n = 0$. This equation is solved with a multigrid algorithm. Finally,

$$V^{n+1} = V^* + \delta t \nabla \Phi, \quad (8)$$

$$P^{n+1} = P^n + \frac{3}{2} \Phi. \quad (9)$$

For the spatial discretization we use the classical staggered grid arrangement, where the viscous and convective terms are treated by finite centered differences.

2.4. Computational domain

The overall domain is rectangular, with $N_x \times N_z$ rectangular cells through which the boundaries can move freely. The grid can be refined in areas of particular interest. The immersed boundaries are taken into account through a phase function equal to 0 or 1 for fluid or structure cells respectively, and velocities are implicitly set to zero in the structure cells. This phase function is rendered time-dependent to admit boundary deformation.

The design and temporal deformation of the flow boundaries follow the choices made in [7]: the glottal surface is described by two sinusoidal curves connected by a tangent straight line, joining two points of coordinates $(x_u, H_u(t))$ and $(x_d, H_d(t))$. These two points play the role of the punctual masses in lumped vocal fold models. When vocal fold oscillations are forced, the upstream (H_u) and downstream (H_d) margins of the line are animated with a vertical movement which emulates that of the masses supporting the glottal surface in conventional two-mass models.

$$H_u(t) = H_0 + A_u \cos(2\pi f t + \phi), \quad (10)$$

$$H_d(t) = H_0 + A_d \cos(2\pi f t). \quad (11)$$

H_0 is the mean height value, A_u and A_d the oscillation amplitudes, f the oscillation frequency, ϕ the phase lag between the upstream and downstream margin locations, and t is time. The upstream and downstream margins stand for the inferior and superior margins of the vocal folds. The fluid lies between the symmetrically varying vocal-fold profiles. To avoid vocal fold collision, the minimal diameter of the constriction is kept greater than zero. Notice that the phase lag ϕ is considered, in forced vocal-fold numerical models, an external constant parameter. In terms of the acoustic output, ϕ is related to the speed quotient, a parameter which is used in acoustics and speech synthesis to parametrize the skewness of the glottal pulse [22]. Throughout this work, ϕ will be kept constant.

The glottal half-angle θ , which will be used in the next section to describe the instantaneous angular configuration of the channel, is defined as the angle of tangent $(H_d - H_u)/(x_d - x_u)$.

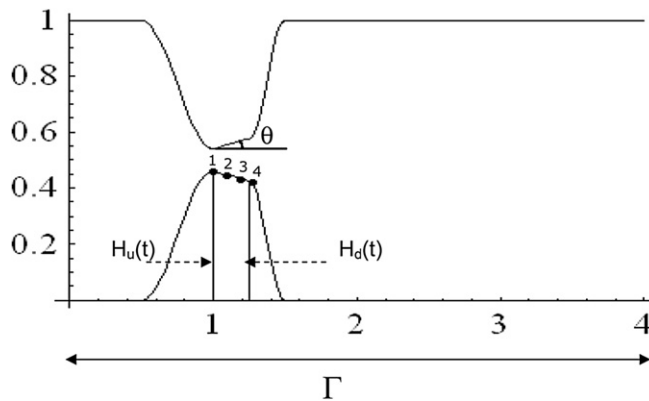


Fig. 1. Glottal channel geometry.

2.5. Numerical parameters and nominal conditions

The computations were performed on a grid with either 512×128 cells (low spatial resolution) or 1024×256 cells (high spatial resolution) in the axial and transversal directions respectively. Calculations can be performed from a zero initial condition, as well as from previously computed solutions. The dimensionless computational domain is a rectangle of unit height and adjustable length (aspect ratio) Γ .

From [7], we adopt the following nominal conditions: half-channel height $h = 1.25$ cm, $x_u = 6$ cm, $x_d = 6.5$ cm, $A_u = 0.75$ mm, $A_d = 0.90$ mm, $\phi = 60^\circ$, $f = 1$ (100 Hz for $Re_h = 1000$). The dimensionless time step was in general kept equal to 10^{-4} (0.001 msec if $Re_h = 1000$). The inlet duct can be shortened without significant changes in flow behavior.

2.6. Numerical scenarios

As advanced in the introduction, we consider a number of different scenarios in order to illustrate the variability of flow dynamics within the constriction. In a first stage, we perform numerical simulations setting a constant inlet flow but assuming flow conditions such that a low spatial resolution is required. Namely, the constriction flaps but maintains a minimal diameter of around 10% H and we set $Re = 500$, which is low with respect to more realistic values ($Re = 2000$). Under these *mild* conditions, we study flow behavior within the constriction in four cases: static configurations, oscillating downstream height, oscillating upstream height and finally both heights oscillating simultaneously. Finally, we turn to a more realistic scenario, in which we use a higher Reynolds number ($Re = 2000$) and an almost closing constriction (with a minimal diameter of 0.8% H), simultaneously relaxing the condition on the entrance flow, which is no longer constant, but synchronized with wall motion, as in [7].

3. Results

3.1. The static case

Let us first present some general considerations regarding flow behavior past the constriction when vocal folds are kept still. Visualizations of flow evolution (axial velocity) for the nominal conditions and $Re = 500$ in a divergent configuration is shown in Fig. 2. A pair of vortices is generated during the transient, which disappear after their passage. This is characteristic of a convective instability in an open flow. When the instability leaves the channel ($t \sim 0.6$), a steady state seems to be achieved, which is symmetric with respect to the channel axis. However, as time progresses (towards $t \sim 1.6$), the jet loses stability, oscillates and breaks the flow axial symmetry. The vortices propagate downstream, resulting in a sort of wavy core flow with lateral oscillations in the downstream direction. Studies of the fully developed flow show that about $Re > 100$ this flow becomes unsteady, a result that is close to the behavior found in [21] for a wavy channel. For higher Reynolds numbers (as is the case for glottal airflow), fluid flow

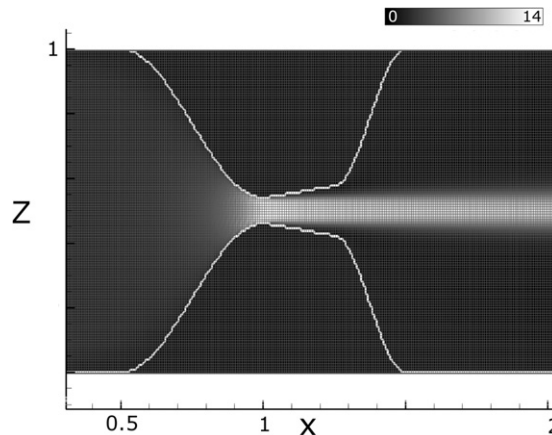


Fig. 2. The constriction of the glottal channel: contour values for U at early times of the numerical simulation.

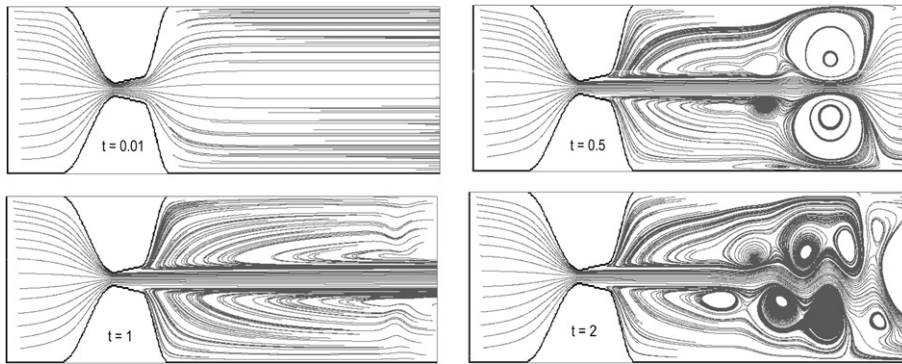


Fig. 3. The instantaneous streamlines at $t = 0.01, 0.5, 1, 2$ for the nominal conditions.

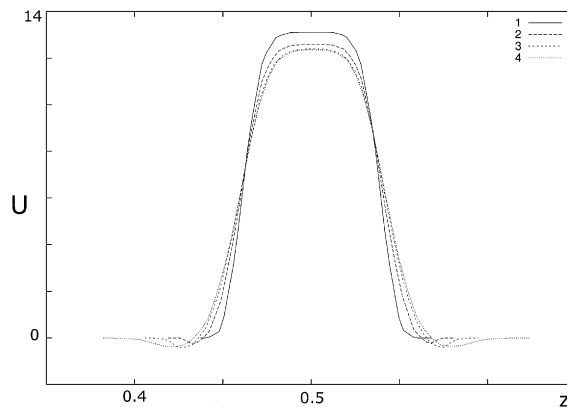


Fig. 4. Glottal velocity profiles at $t = 2.5$ for a few selected x -grid locations (labeled in Fig. 1).

becomes hence intrinsically unstable to a sinusoidal mode. These results, shown in Fig. 3, are also in line with the observations reported in [23]. Simulations show that the instability sets in more rapidly if the channel's configuration is convergent and is even more rapidly damped if the walls are parallel.

The velocity profiles across the glottis at a few selected cross sections in dimensionless units are shown in Fig. 4, in an overlaid format with the grid numbers that refer to locations in Fig. 1. The velocity magnitudes increase at the minimum cross-sectional area and vanish on the wall due to the no-slip condition. Near the wall, the velocity gradient ($\partial U / \partial Z$) becomes negative, which indicates the flow reversal occurring at flow separation. Such profiles are not significantly sensitive to the flow characteristics further downstream.

Let us stress that the unsteadiness of the fully developed flow past the constriction does not significantly affect the velocity or pressure profiles within the constriction. This property of the static case is also observed by the separation point, which moves upstream into the constriction at early times in the simulation till it reaches a location which is maintained thereafter, regardless of the instabilities further downstream. When the constriction flaps, the unsteadiness of the airflow past the constriction will remain quite superfluous regarding the intraglottal phenomena that we will inspect in the next sections. Within the constriction, flow dynamics during the flapping motion will be interesting and nontrivial, but will remain sensitive to flow behavior at the entrance of the glottal constriction rather than to the transglottal instabilities discussed above.

3.2. The dynamic case

3.2.1. Constant inlet flow

In this section, we present first a series of numerical experiments, all of which are carried out with a constant velocity profile at the inlet of the channel, compatible with a constant subglottal pressure. This boundary condition has been married with flow conditions allowing for numerical stability with a relatively low spatial resolution. These

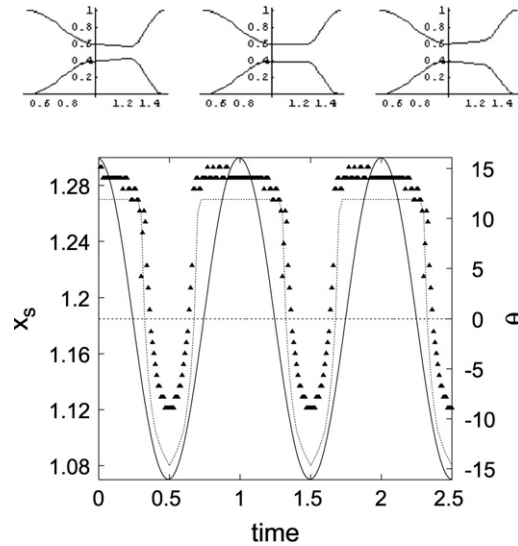


Fig. 5. Numerical simulations for a flapping downstream margin $H_d(t)$, constant inlet flow and $Re = 500$. Above, convergent ($\theta > 0$), parallel ($\theta = 0$) and divergent ($\theta < 0$) vocal fold configurations. Below, the plot shows the evolution of the glottal half-angle θ in degrees for the flapping constriction (full line), overlaid with the separation locations $x_s(t)$ during the flapping motion (triangles) and the $x_s(\theta)$ (pointed line) corresponding to the static case for a fixed half-angle θ .

conditions involve avoiding severe constrictions during the closing phase of the cycle (minimal aperture of 13.6% H) and moderating the intensity of the incoming velocity (i.e. working with $Re = 500$). It is clear that a highly constricted channel together with a constant incoming flow falls out with the incompressibility condition and therefore strains numerical stability for a given spatial resolution.

In this scenario, we perform numerical simulations to study the behavior of the boundary of the effective flow tube (delimited by a zero axial velocity) relative to motion of the two margins of the glottis. Let x_s be the horizontal coordinate where the boundary of the flow tube intersects the lateral wall representing one of the vocal folds (axial flow asymmetries, which can be quantified in our numerical experiments, are not considered in this study).

In the first dynamic case we let only the downstream margin oscillate (H_u is constant, i.e. $A_u = 0$). A sequence of convergent-parallel-divergent configurations for this case is shown in Fig. 5, together with the evolution of x_s resulting from our numerical simulations.

From Fig. 5, we can see that, when only the upper vocal fold margin moves, flow separation is in phase with wall motion. When the constriction is convergent ($\theta > 0$), flow detaches at the end of the constriction (around $x_s = 1.28$). Instead, when the glottis is divergent, x_s moves, first inward and then outward, within the glottal constriction. Except for the bounds of the excursion of x_s into the constriction, which has the time to reach farther locations if the numerical experiment is static (pointed line), we can assert that flow separation evolves in a quasi-static fashion when the only source of flow modulation is the motion of the downstream margin.

The result changes if it is the upstream margin that moves instead (see Fig. 6). Here, H_d is fixed and $H_u(t)$ is a sinusoidal function, having an amplitude A_u which is slightly smaller than A_d in the previous case. Notice that now, flow separation is retarded with respect to wall motion with a time lag of about 20% of the time period of the glottal cycle. Note also that the amplitude of the excursion of $x_s(t)$ is yet smaller than it was in the previous case (recall that the A_u is also smaller here). The appearance of a delay between wall motion and flow separation is in line with the findings reported in [7]. In particular, notice that the mobility of flow separation persists into the convergent phase of the cycle, and that this occurs also in the absence of the fluctuating tracheal flow used in [7]. For the data shown in Fig. 6, the only source of flow modulation which can account for this delay is the motion of the upstream margin (since we have set the incoming flow constant). Therefore, if delay depends on the rate of change of the flow as suggested in [7], it is strongly dependent on *where* in the channel, flow variations are most significant. In other words, if flow modulation is only introduced by wall motion about the exit of the constriction, the delay between flow separation and wall motion is no longer appreciable. This suggests that specification of a varying tracheal input velocity should not be innocuous in the description of intraglottal flow dynamics. This boundary condition probably deserves further

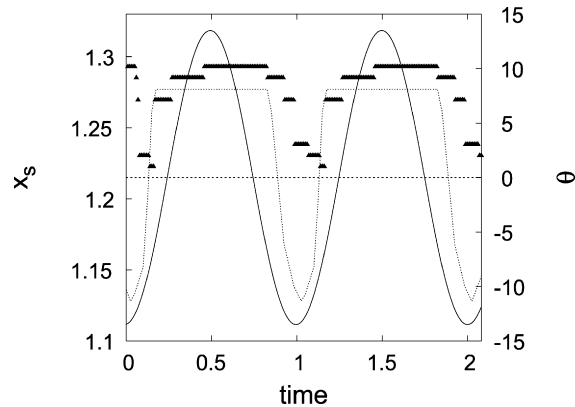


Fig. 6. Numerical data obtained for an oscillating upstream margin $H_u(t)$ with amplitude A_u (smaller than A_d in Fig. 5), constant inlet flow and $Re = 500$.

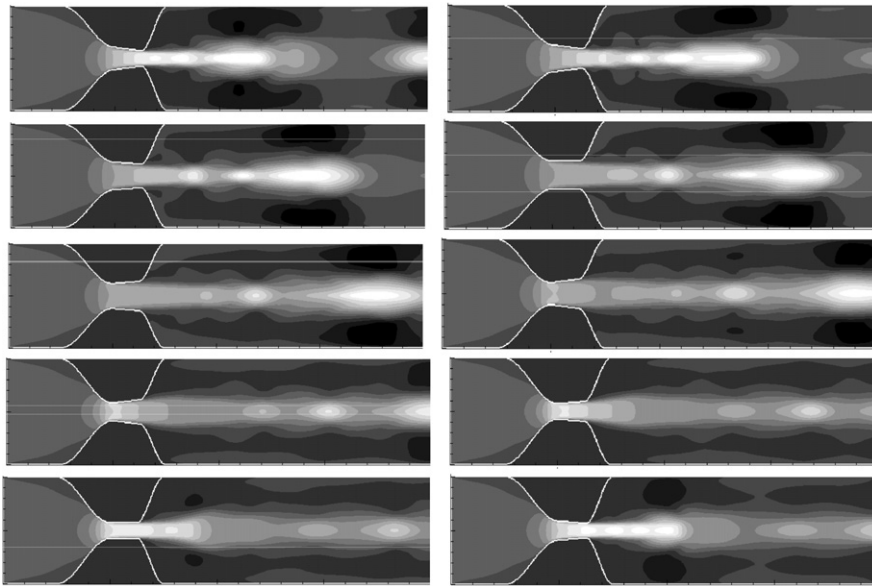


Fig. 7. Glottal axial velocity in the glottis for different times covering a complete glottal cycle for $Re = 500$.

attention and research. Because real glottal wall motion involves simultaneous displacements of *both* margins (the lower part of the folds oscillating with more amplitude than the upper part), these results also suggest that a time-lag between flow separation and wall motion is likely to occur in real glottal flow.

Let us now consider both margins moving. When the glottal cycle begins, subglottal pressure moves apart the vocal folds which adopt a convergent configuration (with H_d minimal). As time progresses, flow passes through the constriction, opening the downstream margin of the vocal folds, until a parallel configuration is reached. Next, the boundary adopts a divergent configuration and the vocal folds start closing. A parallel (but narrower) configuration is again produced during the closing phase, and finally both folds encounter their initial position. Flow visualizations for different configurations within a glottal cycle in which both margins move (according to Eq. (10)) are shown in Fig. 7.

If we inspect the evolution of flow detachment from the walls during the oscillations of both margins in these numerical experiments (Fig. 8), we find that $x_s(t)$ remains invariant near the constriction exit regardless of the angular configuration of the walls during about 70% of the cycle. Mobility is confined to the fully open phase of the cycle where the glottal constriction is larger ($d\theta/dt < 0$). Flow behavior is not quasi-static: the excursion of x_s into the constriction is twice as quick and short as it is for a steady flow. Sensitivity of flow separation to the constriction's largeness seems to be overriding the dependence on the angular configuration in these simulations.

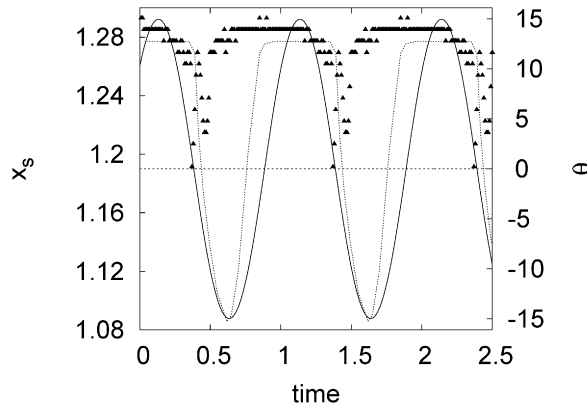


Fig. 8. Numerical data obtained when both margins oscillate with a fixed phase lag of $\phi = \pi/3$.

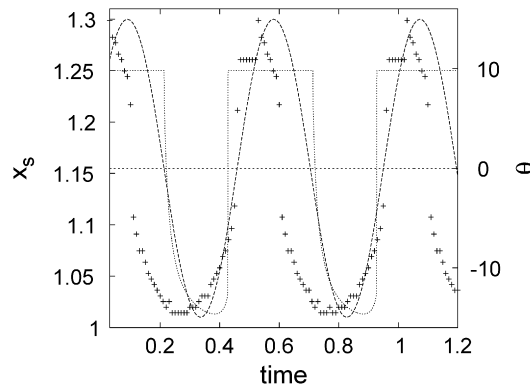


Fig. 9. Flow separation locations (points) and glottal half-angles (full line) as a function of time in dimensionless units corresponding to $Re = 2000$ and modulated inflow. The dashed line represents the separation points predicted by the quasi-static geometrical criterion, used in some vocal-fold models (Liljencrant's condition).

To sum up, results in this section confirm that the amplitude and rate of change of the volume flow have a strong influence on how flow detaches from the vocal fold walls within the flapping glottal constriction. Flow separation dynamics is quasi-static only if flow modulation is not appreciable at the entry of the glottal constriction. Otherwise, intraglottal flow dynamics should be expected to be nontrivial.

In order to study flow separation dynamics in a channel with moving boundaries conforming to actual vocal-fold motion, we must quit this provisional scenario and turn to a highly constricted channel traversed by a faster flow.

3.2.2. Time-dependent inlet flow

Let us now consider the case of a faster flow ($Re = 2000$) and a severe constriction that reaches a minimal diameter as small as 0.8% of the channel's height. These settings require higher spatial resolution, mesh refinement around the axis, and a specification of a time-dependent tracheal flow reconciling flow incompressibility with an almost complete vocal-fold closure. We will adopt the decisions made in [7], where flow is synchronized in a sinusoidal fashion with the motion of the glottal walls. The inlet tracheal mean velocity $U_T(t)$ is set such that at the beginning of the cycle, which occurs when the downstream margin is at its minimal diameter, the velocity is zero. The inlet velocity is thus in phase with the diameter of the upper part of the vocal folds, i.e. $U_T(t) = 0.5 U_0(1 - \cos(2\pi ft))$, where U_0 is determined by the choice of Re .

Let us recall, even if this case is geometrically more realistic, that in real phonation, the incoming flow is not independently reduced during the closing gesture as it is here (it is reduced through flow resistance as the glottal diameter diminishes), and that this condition will tend to emphasize flow separation. In this scenario, we have again inspected the temporal evolution of the detachment of the effective flow tube from the vocal-fold walls. Results are shown in Fig. 9.

The first remark to be made is that, unlike the cases discussed above, $x_s(t)$ now varies almost permanently and only stays fixed near the glottal exit during a small fraction of the cycle (10% here, against 70% in Fig. 8). The excursion of x_s into the constriction is also smoother and amplified with respect to the cases discussed before. Surprisingly, the minimal value of $x_s(t)$ is reached *before* the vocal folds attain their most divergent configuration, with a time-lag of about 20% of the glottal cycle, of the same order as the time-lag observed in Fig. 8. Compared to the values of $x_s(t)$ predicted by the geometrical (quasi-static) criterion invoked in some vocal fold models, it is clear that the dynamic case yields different results, which are counterintuitive if one attempts to explain the intraglottal phenomena from quasi-static observations. In view of our numerical simulations, it is reasonable to assert that flow separation dynamics is far from being either a secondary or a quasi-static effect. The specification of tracheal flow near the glottal entry is certainly a determinant boundary condition in numerical simulations, which should be carefully studied before the validity of numerical experiments to describe realistic glottal flow separation dynamics can be assessed.

4. Conclusions

In this work, we present a series of numerical simulations of airflow in a channel with a constriction that is animated with the typical flapping motion which is characteristic of vocal-fold oscillations. The implemented numerical method integrates the unsteady Navier–Stokes equations, based on an immersed boundary approach with a fixed Cartesian grid, in contrast to the more conventional body-fitted grids used in the literature. The proposed approach does not need remeshing to conform with wall motion, having hence a lower computational cost and a discretization scheme of inalterd accuracy, due to the rectangular grid. The glottal wall is introduced as a freely-moving solid through the grid. Our model is different from previous studies in its numerical scheme, but also in being able to combine a steady entrance flow with wall motion. Data extraction processes are integrated to the numerical simulation, so that the dynamic development of significant variables (allowing to detect, for instance, where the flow tube detaches from the walls) can be directly evaluated. An exhaustive study of flow variables (such as intraglottal pressure or wall pressure) will be addressed in a separate work. Last but not least, we do not include fluid-structure interaction.

The numerical experiments that we present are designed to illustrate the variability in the mobility of flow detachment from the moving rigid walls. We show, first, a few results concerning flow instabilities past the constriction, to proceed afterwards with simulations of the flapping constriction for a constant inlet flow (mimicking constant subglottal pressure) and finally for a faster flow with a specified time-dependent tracheal velocity synchronized with wall motion.

The constant inflow boundary condition is not easy to match with severe glottal closures due to the incompressibility condition. It has been therefore applied together with flow and geometric conditions (moderate incoming flow and large minimal glottal aperture) that preserve numerical stability when a relatively low spatial resolution is used. Vocal-fold flapping motion means setting the upper and lower parts of the vocal folds to oscillate with a mutual (constant) phase lag. In this context, we inspect flow behavior in three distinct situations: oscillating downstream margin, oscillating upstream margin, and both margins oscillating simultaneously with a fixed phase lag.

Our results suggest the following:

- Flow instabilities past the constriction prevent flow solutions from achieving a steady state. However, the unsteady character of the flow past the constriction does not seem to affect flow behavior within the glottal constriction, that is periodical with the frequency of the forced oscillations.
- Increasing the average flow systematically pushes the excursion of the separation point further upstream in the constriction.
- Similarly, we find that the most upstream locations are reached when the half-glottal angle is large and negative (divergent), for different flow and boundary conditions.
- In general, inward wall motion seems to squeeze the separation point downstream, unless it is already located at the exit.
- Quasi-static intraglottal flow dynamics can be detected for a constant inlet flow, if flow modulation is introduced exclusively by the motion of the downstream margin. Otherwise, a time lag between flow separation and wall motion is systematically observed.

- The time lag between wall motion and flow separation is not necessarily negative (flow separation can occur before the most divergent angular configuration is reached); its length depends on the rate of change of the flow, but also on *where* in the channel this rate of change is most significant.
- Mobility of the flow separation point is emphasized when a time-dependent tracheal flow is prescribed. In particular, for a tracheal flow that is in phase with the upper glottal margin, flow separation occurs at different positions all over the cycle, holding up near the exit only during a small fraction of the cycle.
- The geometrical criterion for flow separation used in some vocal fold models (Liljencrant's condition) is not a good approximation of the results we obtain in our numerical experiments.

In fact, our results show that it is decidedly not easy to identify the conditions that systematically control reversals of intraglottal flow separation mobility. Furthermore, the examples that we present prove that the picture of inertial forces retarding flow effects with respect to wall motion might not be pervasive either. These numerical experiments attest both, the importance and the diversity of flow separation dynamics in a constriction sharing some of the features which are characteristic of vibrating vocal folds. They also suggest that quasi-static or geometric criteria should be abandoned in favor of a dynamical criterion. This issue calls for further research under conditions being both, numerically robust and realistic enough to throw light on the physics of glottal airflow, and to construct more suitable models of fluid-mechanical effects within the glottis. A close examination of these questions makes part of our work in progress.

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