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(Received 25 August 2025; accepted 13 January 2026; published 27 February 2026)

The atomization of airway mucus during speech represents a primary mechanism for respiratory aerosol generation, which is central to airborne pathogen transmission. While extensive research has examined single-phase phonation aerodynamics, the multiphase dynamics of droplet formation has received less attention. This study aims to fill this critical gap by presenting a multiphase computational fluid dynamics (CFD) model of the human vocal folds. The dynamic formation and subsequent breakup of the mucus bridge are simulated using a volume-of-fluid method coupled with an analytical model describing the time-dependent shape of the glottal orifice. The investigation reveals that initial, small-scale stochastic perturbations on the mucus surface are essential for triggering a physically realistic, multimodal breakup, in contrast to the delayed, symmetric rupture observed in idealized smooth-film simulations. During vocal-folds opening, the bridge thickness exhibits exponential decay rupturing after 2.32 ms at a dynamic aspect ratio of approximately 20, a value significantly exceeding the static stability limit. Furthermore, a parametric study reveals that increasing transglottal pressure from 491 Pa to 1472 Pa, corresponding to louder phonation, accelerates the thinning process and leads to earlier initial rupture after 2.04 ms. This work provides fundamental insights into the precursors of respiratory aerosol generation

DOI: [10.1103/qq23-l918](https://doi.org/10.1103/qq23-l918)**I. INTRODUCTION**

The atomization of airway-lining fluid during speech is a primary source of respiratory droplets. Particles with diameters of a few micrometers or less form aerosols that remain suspended in

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ambient air, facilitating the airborne transmission of pathogens such as influenza, measles, SARS-CoV-2, and tuberculosis [1].

It has been shown that during quiet breathing, submicron droplets are predominantly generated in the lower airways by the rupture of liquid plugs that temporarily occlude the alveoli [2]. In contrast, larger droplets (on the order of micrometers) are primarily produced during vocalization, such as speaking, singing, or shouting, from the atomization of fluids lining the upper airways [3–5]. The fluid properties and flow geometries of the upper airways, however, differ substantially from those of the lower airways. Consequently, insights into aerosol generation in the alveoli cannot be readily extrapolated to the mechanisms occurring in the vocal tract.

Experimental investigations have begun to characterize these multiphase phenomena. Fritzsche *et al.* [4] used an actively oscillating silicone model of the vocal folds to mimic phonation, observing the formation of liquid filaments and films that subsequently ruptured into droplets, similar to mechanisms seen at the lips [6]. Their work showed that the resulting bimodal-droplet size distribution was in good agreement with data from human subjects [5], identifying the core macroscopic mechanisms.

To gain deeper insights, numerical modeling using computational fluid dynamics (CFD) has been widely applied to study phonation. A recent review by Döllinger *et al.* [7] categorizes these efforts based on their physical complexity, including

- (i) CFD models with static or prescribed mesh motion and optionally with acoustic analogies,
- (ii) Fluid-structure interaction (FSI) models based on Navier-Stokes or Bernoulli's equations without acoustics, and
- (iii) Fluid-structure-acoustic interaction (FSAI) models.

The first category, pure CFD models, has been used to investigate fundamental aerodynamic phenomena. Static models, employing custom geometries [8], simplified half cylinders [9], or the well-established Scherer M5 model [10], have been applied to study the effect of glottal angle on pressure distribution [11,12] and the influence of the ventricular folds [13,14]. Models with prescribed kinematics have clarified the evolution of the glottal jet and vortex shedding [15–17] and detected acoustic sources [18]. The more complex FSI models compute the vocal-fold motion as a dynamic response to aerodynamic forces, enabling the study of self-sustained oscillations and energy-transfer mechanisms [19–21]. The most comprehensive FSAI models couple fluid, structural, and acoustic fields to investigate phenomena such as vocal-tract resonance and its feedback on vocal-fold vibration [22,23].

However, these established simulation approaches are almost exclusively based on single-phase flow assumptions, which is insufficient for modeling atomization. Some studies have, however, considered aerosol transport and dispersion using Lagrangian particles [24–26] or mucus clearance during coughing [27–29]. Furthermore, Kant *et al.* [30] investigated bioaerosol generation during violent respiratory maneuvers, such as coughing or sneezing. However, to the best of the authors' knowledge, the multiphase dynamics of mucus-bridge formation and primary breakup during vocal-fold opening have not been simulated. This study addresses this gap by presenting a multiphase CFD model focused specifically on these precursor events to aerosol generation.

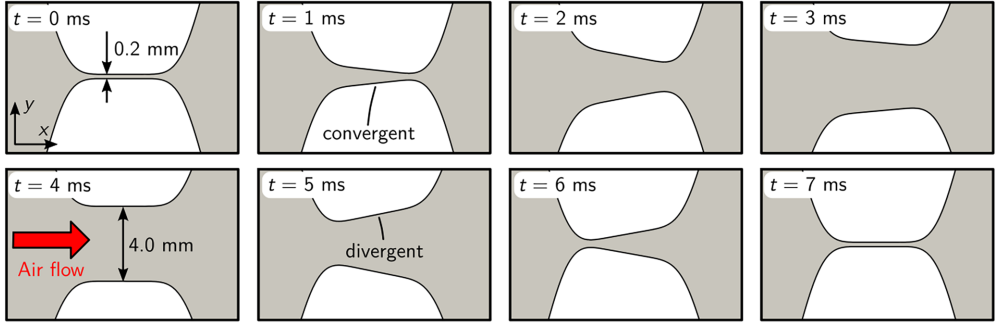
II. NUMERICAL MODEL

The mucus-bridge breakup between the vocal folds was simulated with the solver *interIsoFoam* of the open-source finite-volume method (FVM) CFD library *OpenFOAM* v2412 [31]. The simulations were performed with up to 2560 CPU cores per case and 2 weeks of execution time at the Center for Information Services and High Performance Computing (ZIH) at TU Dresden, Germany.

A. Geometry and mesh

The geometry of the computational domain was adopted from an analytical model of Zhao *et al.* [32] (see grey lines in Fig. 2, left). They employed a time-dependent shape of the glottal orifice

(a) Longitudinal view (x-y plane)



(b) Cross-sectional view (y-z plane)

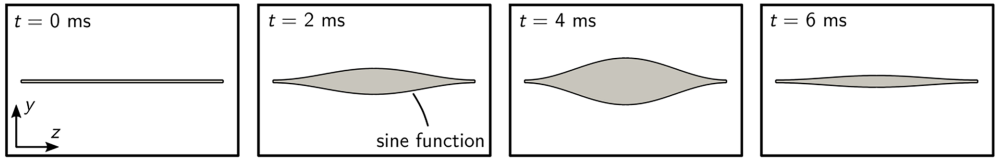


FIG. 1. Time-dependent shape of the glottal orifice over one period of $T = 7$ ms ($f_0 = 142$ Hz): (a) The longitudinal view (x-y plane) illustrates the convergent-divergent motion characteristic of the mucosal wave, moving from a minimum opening of 0.2 mm to a maximum of 4.0 mm, and (b) the cross-sectional view (y-z plane) shows the sinusoidal shape applied along the anterior-posterior length of the vocal folds.

$y_g(x, z, t)$ consisting of three terms:

$$y_g(x, z, t) = y_0(x) + y_t(x, t) \cdot \Delta y_z(z). \quad (1)$$

The first term defines the static shape of the glottis orifice $y_0(x)$ represented by a hyperbolic tangent as follows:

$$y_0(x) = \frac{D_0 - D_{\min}}{4} \tanh(s(x)) + \frac{D_0 + D_{\min}}{4} \quad (2)$$

with the dimensionless parameter s defined as

$$s(x) = \frac{b|x|}{D_{\max}} - \frac{b D_{\max}}{|x|}, \quad (3)$$

the coordinate in streamwise direction x , the maximum and minimum glottis opening of $D_{\max} = 4$ mm and $D_{\min} = 0.2$ mm, respectively, the duct height of $D_0 = 20$ mm, and constant $b = 1.4$. This static shape can be seen in Fig. 1(a) at $t = 0$ ms.

The second term $y_t(x, t)$ stands for the oscillatory motion of the vocal folds, creating convergent, divergent, closed, or fully opened configurations:

$$\begin{aligned} y_t(x, t) = & \frac{1}{2}[1 - \tanh(s(x))] \cdot \beta_1(t)(x + c D_{\max}) \\ & - \frac{1}{2}[1 - \tanh(s(x))] \cdot \beta_2(t)(x - c D_{\max}) \end{aligned} \quad (4)$$

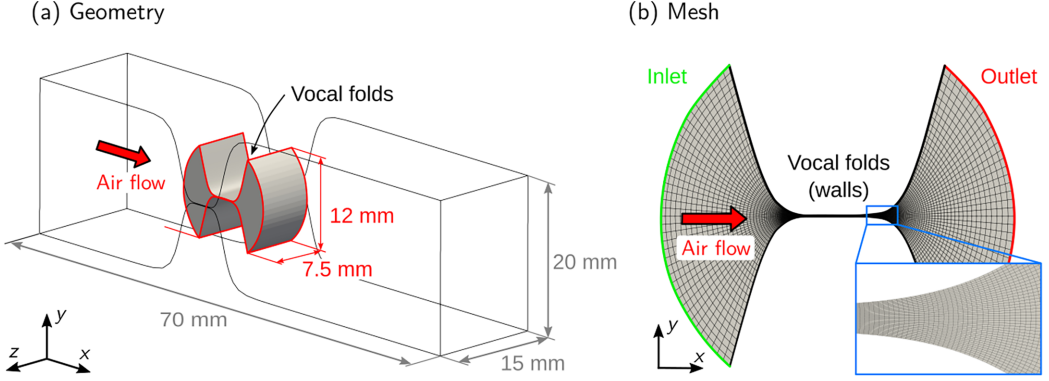


FIG. 2. Computational domain and mesh: (a) Reduced domain (red) adapted from Zhao *et al.* [32], and (b) the structured mesh provides high resolution at the glottal orifice and coarsens towards the inlet and outlet boundaries (every fifth line shown).

with constant $c = 0.4$. Time-dependent variables β_1 and β_2 are functions during one cycle of the glottis motion that determine the actual timing of any given instantaneous shape:

$$\beta_1(t) = \begin{cases} 0 & \text{for } t \leq \frac{1}{9}T, \\ 0.224 \left\{ 1 - \cos \left[\frac{9\pi}{4} \left(\frac{t}{T} - \frac{1}{9} \right) \right] \right\} & \text{for } \frac{1}{9}T < t \leq \frac{5}{9}T, \\ 0.488 & \text{for } \frac{5}{9}T < t \leq \frac{6}{9}T, \\ 0.224 \left\{ 1 + \cos \left[3\pi \left(\frac{t}{T} - \frac{6}{9} \right) \right] \right\} & \text{for } \frac{6}{9}T < t \leq T, \end{cases} \quad (5)$$

$$\beta_2(t) = \beta_1(t + T/9), \quad (6)$$

based on period length $T = 1/f_0$ with f_0 being the vocal-folds oscillation frequency of $f_0 = 142$ Hz. This is visualized in Fig. 1(a) at $t > 0$ ms.

The third term $\Delta y_z(z)$ in Eq. (1) applies a sinusoidal variation to the glottal opening width in anterior-posterior direction:

$$\Delta y_z(z) = \frac{1}{2} \left[\sin \left(2\pi \left(\frac{z}{L} - \frac{1}{4} \right) \right) + 1 \right], \quad (7)$$

with L being the length of the glottis in anterior-posterior direction. The resulting opening and closing of the vocal folds over one period is shown in Fig. 1(b).

Compared to the original geometry from Zhao *et al.* [32], the size of the computational domain for this study is reduced as shown in Fig. 2. This decreases the computational cost considerably and focuses on the glottal region where breakup occurs, assuming far-field effects are negligible for the breakup. While the modeled vocal-fold length is $L = 7.5$ mm, it does not represent pediatric dimensions as the original geometry is based on adult vocal-fold anatomy and the computational domain presented is only a section of it. This simplification is solely to reduce computational cost. Furthermore, this does not affect the breakup mechanism because the sinusoidal-opening profile causes initial rupture to occur at the center of the domain, where the glottal opening and consequently the mucus-bridge stretching is maximum. The computational domain consists of a structured mesh, where three different mesh sizes are created for the mesh-independency study (see Table I). The structured topology of the mesh automatically results in a high mesh resolution at the glottis and a coarser mesh at the inlet and outlet boundaries. The actual cell size varies throughout the simulation as the glottis is opening. For the finest mesh, it is approximately $8 \mu\text{m}$ during the formation and breakup of the mucus bridge.

The time-dependent shape of the glottal orifice, based on Zhao *et al.* [32], is rarely used compared to the well-established Scherer M5 model [10]. However, the former has two main advantages for

TABLE I. Cell count in streamwise, wall-normal, and spanwise direction; total mesh size; and approximate cell size during the formation and breakup of the mucus bridge.

Mesh	Cell count	Mesh size $\times 10^6$	Cell size [μm]
Coarse	$340 \times 72 \times 202$	5.0	20
Medium	$540 \times 116 \times 316$	19.8	13
Fine	$864 \times 184 \times 504$	80.1	8

this study: (i) Ease of implementation due to a simple mathematical description of the vocal-folds surface, and (ii) gentle slopes upstream and downstream of the vocal folds. In particular, the latter advantage allows for a high-quality, structured mesh topology throughout the vocal folds opening and closing. Because an arbitrary Lagrangian-Eulerian (ALE) approach can result in highly distorted cells in the glottis, Scherer's M5 model is often combined with more complex meshing strategies, such as overset mesh method [16,33], or immersed boundary method [34,35].

B. Governing equations

The physics of multiphase flow inside the vocal folds is governed by the three-dimensional, incompressible Navier-Stokes equations

$$\frac{\partial \rho}{\partial t} + \nabla \cdot (\rho \mathbf{u}) = 0, \quad (8)$$

$$\frac{\partial \rho \mathbf{u}}{\partial t} + \nabla \cdot (\rho \mathbf{u} \mathbf{u}) = -\nabla p + \mu \Delta \mathbf{u} + \rho \mathbf{g} + \mathbf{f}_\sigma, \quad (9)$$

with the fluid density ρ , fluid velocity $\mathbf{u}$, static pressure p , dynamic viscosity μ , gravitational acceleration $\mathbf{g}$, and surface tension force $\mathbf{f}_\sigma$. In order to account for the two-phase laminar flow of air and mucus, the model is extended for two incompressible, immiscible fluids using the volume-of-fluid (VoF) method with the `isoAdvector` geometrical interface capturing and advection method developed by Roenby [36] and extended by Scheufler *et al.* [37]. Here, an additional transport equation is solved:

$$\frac{\partial \alpha}{\partial t} + \nabla \cdot (\alpha \mathbf{u}) = 0, \quad (10)$$

with the volume fraction α representing for each cell the fraction of its volume which is occupied by one of the two fluids:

$$\alpha = \begin{cases} 0 & \text{within the air phase} \\ 0 < \alpha < 1 & \text{interface} \\ 1 & \text{within the mucus phase} \end{cases}. \quad (11)$$

A piecewise linear-interface reconstruction algorithm (PLIC) in combination with a reconstructed-distance function (RDF) is used to reconstruct and advect the sharp interface between both fluid phases. This model has already been validated and used for similar multiphase flow simulations [38].

The surface-tension force $\mathbf{f}_\sigma$ in the momentum equations is formulated as volumetric force according to the continuum surface-force (CSF) model of Brackbill *et al.* [39]:

$$\mathbf{f}_\sigma = \sigma k \mathbf{n}, \quad (12)$$

with surface-tension force coefficient σ , free surface curvature k , and the interface normal vector $\mathbf{n}$ calculated as

$$k = -\nabla \cdot \mathbf{n}, \quad \mathbf{n} = \frac{\nabla \alpha}{|\nabla \alpha|}. \quad (13)$$

Material properties are based on experimental values and weighted linearly based on the phase fraction

$$\rho = \rho_1(1 - \alpha) + \rho_2\alpha, \quad (14)$$

$$\mu = \mu_1(1 - \alpha) + \mu_2\alpha, \quad (15)$$

where index 1/2 stands for the phase of air and mucus, respectively.

Air is modeled as a Newtonian fluid at 37 °C and 100 kPa with constant density and viscosity of $\rho_1 = 1.186 \text{ kg/m}^3$ and $\mu_1 = 1.83 \times 10^{-5} \text{ Pa s}$, respectively. However, mucus has strong non-Newtonian fluid behavior. Therefore, viscosity is computed as a function of shear rate $\dot{\gamma}$ by employing the power law model

$$\mu_2 = \begin{cases} \mu_{\min} & \dot{\gamma} > \dot{\gamma}_{\max} \\ k \dot{\gamma}^{n-1} & \dot{\gamma}_{\min} < \dot{\gamma} < \dot{\gamma}_{\max} \\ \mu_{\max} & \dot{\gamma} < \dot{\gamma}_{\min} \end{cases} \quad (16)$$

with lower and upper shear rates $\dot{\gamma}_{\min}$ and $\dot{\gamma}_{\max}$, respectively.

Exact shear-rate-dependent viscosity, surface tension, and mucus-thickness values of human vocal-folds mucus have not been documented yet and values from different regions in the respiratory tract vary depending on research group. In this publication, we take the experimental values from Fritzsche *et al.* [4] with the power-law coefficients of $k = 4.845 \times 10^{-3} \text{ Pa s}^n$ and $n = 0.756$ and lower and upper viscosity limits of $\mu_{\min} = 1 \times 10^{-3} \text{ Pa s}$ and $\mu_{\max} = 17.78 \text{ Pa s}$, respectively. While these values provide a representative model, it is acknowledged that *in vivo* mucus rheology is highly variable, and this represents a key uncertainty in the current model.

The surface-tension coefficient at the air-mucus interface is defined as $\sigma = 51.9 \times 10^{-3} \text{ N/m}$ and density of mucus is set to $\rho_2 = 1000 \text{ kg/m}^3$ based on a study by Ren *et al.* [40] on the mucus clearance effectiveness in the airways. Mucus-layer thickness used in similar CFD studies varies greatly, e.g., from 30 μm used by Paz *et al.* [27], up to 300 μm used by Ren *et al.* [41]. This study utilizes an average mucus layer thickness of 100 μm throughout all simulations, which is sufficient to fill the initial gap between the vocal folds.

Pressure-velocity coupling is performed using the PIMPLE algorithm, a combination of PISO algorithm by Issa [42] and SIMPLE algorithm by Patankar and Spalding [43]. A single outer correction and 3 inner-corrector steps are performed with no under-relaxation. The convective term in the momentum equations is discretized with a second-order van Leer scheme [44] based on the total variation diminishing (TVD) method [45] and gradients are calculated using least squares. The remaining terms (Laplacian and interpolation) are discretized with second-order Gauss linear schemes. Temporal discretization is second-order Crank Nicolson with a blending coefficient between first order and second order of 0.9, which results in the best compromise between accuracy and stability. The time-step size is automatically adjusted to achieve a constant Courant number of 0.1. A preconditioned conjugate gradient (PCG) solver with a final tolerance of 10^{-7} is used for pressure and a preconditioned biconjugate gradient (PBiCGStab) solver with a final tolerance of 10^{-6} for velocity, respectively.

Preliminary results have shown that the maximum air velocity shortly after mucus-bridge rupture is approximately 20 m/s at a glottal opening width of 2.0 mm. This translates into a Reynolds number of about 2100 rendering the flow laminar. Based on the glottal frequency of f_0 , it takes 7 ms for a complete period of opening and closing of the vocal folds. Since the breakup of the mucus bridge already occurs during the opening phase, only 3.2 ms are required to capture this effect, though.

C. Boundary and initial conditions

The flow is driven by the transglottal pressure difference, defined as the pressure difference between inlet and outlet. However, physiological pressure levels used in CFD studies vary greatly,

e.g., from 300 Pa used by Zörner *et al.* [17] up to 3300 Pa used by Sadeghi *et al.* [16]. In this study, the pressure difference is based on a study of Scherer *et al.* [11], which used values in the range of 294–2453 Pa. For the baseline simulation, the total pressure at the inlet is set to 491 Pa with a static pressure at the outlet of 0 Pa. In order to study the effect of the wide range of different pressure levels, two additional simulations are performed with a total inlet pressure of 981 Pa and 1472 Pa, respectively. The moving walls are considered no slip and perfectly wetting (i.e., wetting angle of 0°).

Under physiological conditions, the mucus layer exhibits natural irregularities due to microscale variations in mucus composition, epithelial surface roughness, nonuniform airflow patterns during vocal-fold oscillation, or coughing [28]. To capture this realistic heterogeneity, initial surface perturbations are introduced to avoid artificially symmetric breakup patterns that would occur in idealized smooth simulations. For these perturbations, a stochastic wave-spectrum approach was utilized to directly modify the volume fraction field in interfacial cells. After identifying the cells at the air-mucus interface (where $0 < \alpha < 1$), a superposition of $n = 20$ wave modes was used to calculate a local dimensionless perturbation $\delta_\alpha(\mathbf{x})$ for each interfacial cell:

$$\delta_\alpha(\mathbf{x}) = \sum_{i=1}^n A_i \cos(\mathbf{k}_i \cdot \mathbf{x} + \varphi_i). \quad (17)$$

Here, $\mathbf{x}$ is the cell-center coordinate vector, $\mathbf{k}_i$ the wave vector for mode i , φ_i a random phase angle uniformly distributed in $[0, 2\pi]$, and A_i the dimensionless amplitude contribution of mode i .

Each wave vector $\mathbf{k}_i$ is generated by first sampling its wave-number magnitude k_i and then assigning it a random orientation. The magnitude k_i for each mode i is sampled from a uniform distribution within a range around the target wave number $k_0 = 22\,500 \text{ m}^{-1}$:

$$k_i \sim \mathcal{U}[k_0 - 3\sigma_k, k_0 + 3\sigma_k], \quad (18)$$

where $\sigma_k = 0.05 k_0$ is the spectral width.

The orientation of each wave vector $\mathbf{k}_i$ is obtained by uniformly sampling spherical coordinates. The azimuthal angle is sampled uniformly as $\theta_i \sim \mathcal{U}[0, 2\pi]$. To ensure an isotropic distribution and avoid clustering near the poles, the polar angle ψ_i is derived by sampling its cosine uniformly: $u_i \sim \mathcal{U}[-1, 1]$, such that $\psi_i = \arccos(u_i)$. The components of $\mathbf{k}_i$ are then defined as

$$\mathbf{k}_i = k_i \begin{pmatrix} \sin(\psi_i) \cos(\theta_i) \\ \sin(\psi_i) \sin(\theta_i) \\ \cos(\psi_i) \end{pmatrix}. \quad (19)$$

The dimensionless amplitude A_i for each mode i is determined by a Gaussian-like distribution

$$A_i = A_{\max} \exp\left(-\frac{(k_i - k_0)^2}{\sigma_k^2}\right), \quad (20)$$

where $A_{\max} = 2$ is the maximum dimensionless amplitude factor. This value was found to be large enough to break the symmetry of the problem and trigger a physically realistic, multimodal breakup, whereas smaller values led to delayed, single-mode breakup.

The calculated perturbation δ_α is then scaled locally by the volume fraction itself using the blending function $f(\alpha) = 4\alpha(1 - \alpha)$ to concentrate disturbances at the air-mucus interface while preserving the average layer thickness:

$$\alpha_{\text{disturbed}} = \alpha + \delta_\alpha(\mathbf{x}) \cdot f(\alpha). \quad (21)$$

Finally, the resulting disturbed-volume fraction $\alpha_{\text{disturbed}}$ is clipped to the physical bounds $[0, 1]$. This approach introduces stochastic variations to the initial volume fraction near the interface, mimicking the effect of real-world disturbances while preserving the specified average mucus-layer thickness and characteristic wavelength distribution.

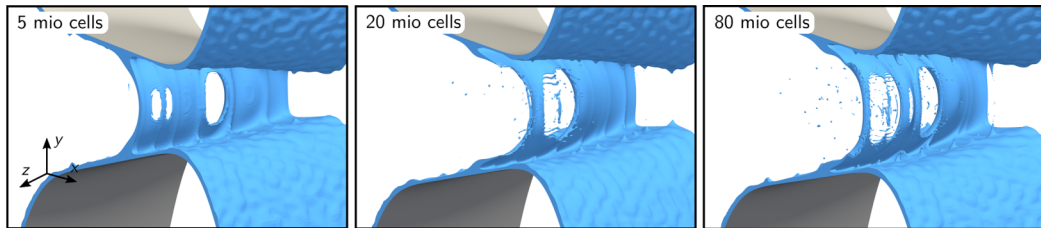


FIG. 3. Mesh dependency of post-rupture structures, shown 0.2 ms after initial breakup. As the mesh is refined from 5 million (left) to 80 million cells (right), the qualitative breakup pattern is consistent, but increased number and finer scale of ligaments and droplets are resolved. Note: Only half of the solution domain is shown for clarity.

III. RESULTS

A. Mesh dependency study

Mesh independence was assessed using three progressively refined meshes with 5×10^6 cells for the coarse mesh, 20×10^6 for the medium mesh, and 80×10^6 for the fine mesh. Two criteria were used to assess mesh independence: the initial rupture time and the qualitative breakup pattern.

The initial rupture times were 1.98 ms for the coarse, 2.18 ms for the medium, and 2.32 ms for the fine mesh, respectively. As the mesh is refined and cell size decreases, a thinner liquid bridge can be simulated, resulting in later breakup times. Figure 3 shows the mucus surface 0.2 ms after initial rupture for the different meshes investigated. Despite different breakup times and the randomness in mucus-layer initialization, the general surface structure and breakup morphology are similar for all three meshes. However, as the mesh is refined, progressively smaller structures are resolved.

These findings indicate that complete mesh independence has not yet been achieved. This is expected given that the reported median droplet diameter for aerosol particles originating from the laryngeal region, particularly during voicing and coughing, is around $1 \mu\text{m}$ [46,47], with some research specifying diameters of $1.7 \mu\text{m}$ for coughing and $2.5 \mu\text{m}$ for speech [5]. These dimensions are much smaller than the cell size during breakup in the numerical model. Based on these expected droplet diameters, a theoretical mesh with a minimum cell size of $2 \mu\text{m}$ during breakup would require a total mesh size of approximately 5.12×10^9 cells assuming uniform refinement in all directions from the current fine mesh.

Given the computational cost of 2 weeks on 2560 cores (Intel Xeon Platinum 8470 at 2.00 GHz with 4 GB RAM per core utilized) for simulating the vocal-fold opening until $t = 3.2$ ms on the finest mesh, further refinement was not feasible with available resources. Therefore, this study focuses on resolving the formation of the mucus bridge and the mechanisms of its initial (primary) breakup, rather than resolving the secondary breakup and characterizing the particle-size distribution of the subsequent aerosol cloud. Despite the mesh dependency, this focus provides valuable insights into the macroscopic precursors of aerosol generation.

B. Mucus-layer initialization

Mucus-layer initialization significantly influences breakup dynamics as initial disturbances are the starting points for the later mucus-bridge breakup. For this reason, a spectrum of initial disturbances is used for a stochastically perturbed mucus layer.

The initialization approach critically determines both breakup timing and morphology. As illustrated in Fig. 4, the smooth initialization results in a delayed breakup ($t = 2.62$ ms) characterized by a single, large, symmetric structure. This pattern fails to reproduce the complex breakup found in literature [4,6]. In contrast, the initially disturbed mucus breaks up earlier at $t = 2.18$ ms and produces a markedly more realistic, fragmented morphology. In both cases, results from the medium-sized mesh are used.

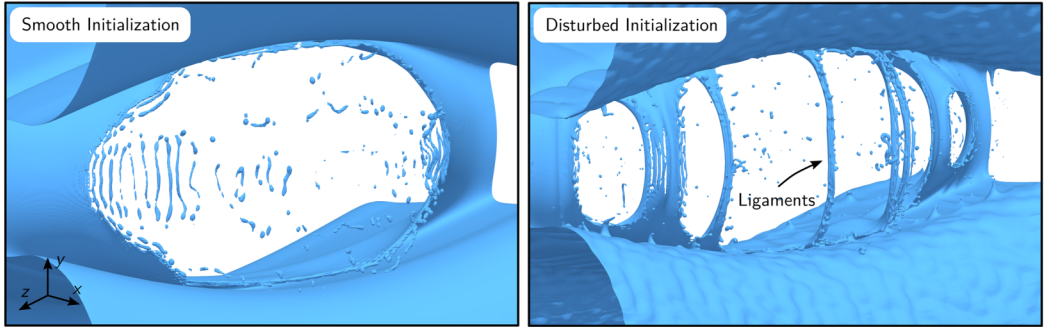


FIG. 4. Influence of initial perturbations on the mucus-breakup morphology at $t = 2.8$ ms. (Left) A smooth initialization leads to a delayed, symmetric breakup forming a single, large structure. (Right) An initially disturbed mucus layer using a stochastic wave spectrum produces a more realistic, fragmented breakup and formation of ligaments.

This difference stems from variations in the initial surface topology. While the smooth case provides a uniform layer, the disturbed initialization creates substantial heterogeneity with local thickness variations as shown in Fig. 5. These initial variations act as seeds for the subsequent instabilities. The key mechanism, visualized in Fig. 6, is that the film preferentially ruptures at locations of initial thickness minima, where the mucus layer is most vulnerable. Conversely, the regions of initial thickness maxima, which contain more fluid, are more resilient and evolve into the ligaments that bridge the newly formed gaps. This results in a direct correlation between the initial small-scale random perturbations and the final large-scale deterministic breakup structure.

The requirement for initial perturbations reflects a limitation of VOF methods in simulating thin-film rupture. When films reach mesh-scale thickness over extended regions simultaneously, as in the smooth case, the interface representation breaks down unphysically, contributing to the symmetric rupture pattern observed in Fig. 4 (left). The stochastic perturbations address this by ensuring the film reaches critical thickness at isolated points, creating realistic perforation sites. However, this approach relies on perturbations persisting throughout the bridge evolution, which may not occur under all conditions. In the future, this problem could be solved more efficiently in two ways: (i) by

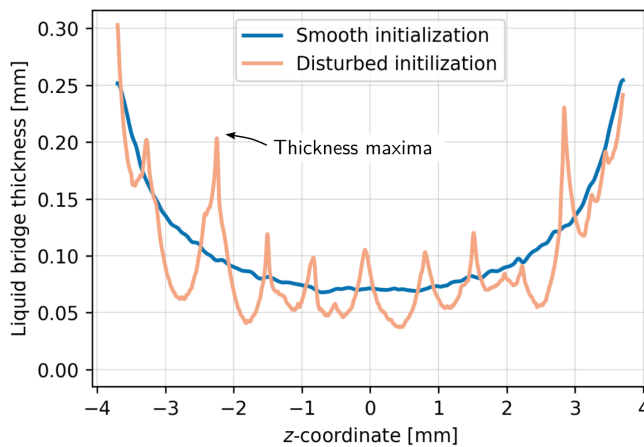


FIG. 5. Impact of initial perturbations on the mucus-bridge thickness along the z axis at $t = 2.0$ ms. The smooth initialization results in minor, symmetric thickness variations. In contrast, the disturbed case creates substantial heterogeneity, with pronounced thickness minima and maxima.

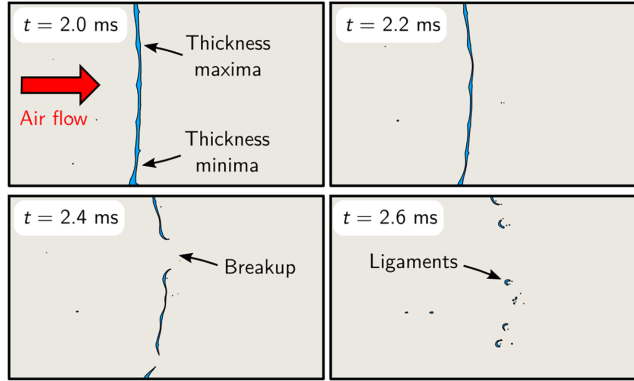


FIG. 6. Visualization of the breakup mechanism seeded by initial perturbations showing a cross section of the mucus bridge. Rupture initiates at a location corresponding to an initial thickness minimum ($t = 2.2$ ms), while the ligaments that form subsequently originate from regions of initial thickness maxima.

controlling film perforation before mesh-resolution limits are reached using methods like manifold death (MD) [48,49], or (ii) maintaining film stability at subgrid scales using reconstruction with 2 planes (R2P) [50,51] or front reconnection [52], or similar approaches.

C. Mucus-bridge formation

The formation and breakup of the mucus bridge is visualized in Fig. 7. Initially, as the vocal folds open, the fixed volume of mucus is stretched. This stage is dominated by prescribed wall motion and surface-tension forces, which cause the mucus to contract into a mucus bridge to minimize its surface area. As the glottal gap widens at $t \approx 1.2$ ms and onward, aerodynamic forces eventually become significant. The pressure difference across the glottis drives an airflow that exerts both shear and pressure forces on the bridge, causing it to deform and accelerate downstream. This dynamic loading rapidly thins the bridge, particularly at the locations of initial weakness, ultimately leading to rupture at $t = 2.32$ ms. The remaining mucus is then transported downstream by the airflow. Note: Since the initial breakup occurs slightly off plane, it cannot be seen in Fig. 7.

The properties of the mucus bridge are further characterized in Fig. 8, which shows the average mucus-bridge thickness, its downstream position relative to the glottis, and kinematic viscosity in the y -normal plane. As the vocal-folds movement starts, an exponential reduction of mucus-bridge thickness can be observed as shown by the nearly linear drop in semilogarithmic scale until the mucus bridge finally ruptures. The position of the mucus bridge is more complex. At first, due to the contraction of mucus within the vocal folds, the mucus bridge is transported downstream. However, after $t \approx 1$ ms, the convection slows down significantly and the mucus bridge becomes thinner due to stretching. Once the mucus bridge becomes thin enough to be significantly affected by the aerodynamic forces (at around $t \approx 1.6$ ms), the mucus bridge accelerates and is quickly convected downstream until it ruptures. The effect of the non-Newtonian behavior of mucus within the mucus bridge can also be observed, in particular at the beginning of the process when the initial mucus layer begins to contract. This results in high shear rates lowering the kinematic viscosity within the mucus bridge by a factor of 2. However, after $t \approx 0.5$ ms, the kinematic viscosity nearly stays constant highlighting that non-Newtonian fluid behavior has only a minor effect after the initial contraction of the mucus bridge.

To analyze the bridge stability, the nondimensional aspect ratio $AR = L/D$ is employed. In this context, $L(t)$ is the minimum opening between the vocal folds, and $D(t)$ is the minimum thickness of the mucus bridge. This definition allows for a simplified comparison to stability theory. For a quasistatically stretched liquid cylinder, the Rayleigh-Plateau instability dictates a breakup threshold

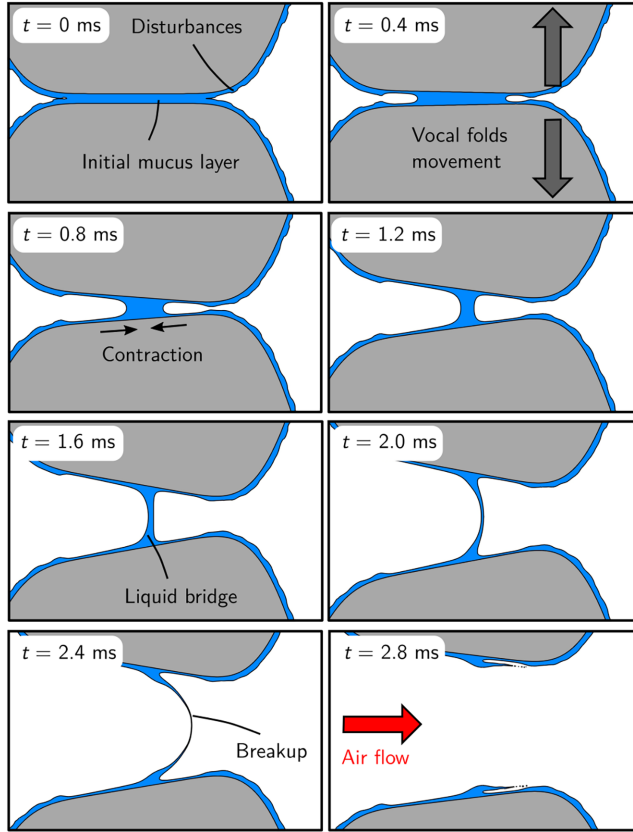


FIG. 7. Time sequence illustrating the multistage process of mucus-bridge formation and breakup. The process begins with the stretching and contraction of the initial mucus layer at $t < 0.8$ ms, dominated by surface tension. As the glottis widens, aerodynamic forces deform the bridge downstream, leading to rapid thinning and eventual rupture.

of $AR > \pi$. The aspect ratio for the mucus bridge presented in Fig. 9 shows a significant deviation from this theory. The rupture event occurs when the aspect ratio reaches approximately 20, which is significantly larger than the static stability limit.

Similar behavior of dynamic liquid bridge rupture has also been reported by Zhuang and Ju [53]. The rapid opening of the vocal folds stretches the bridge at a high strain rate. At this rapid deformation, the system does not have sufficient time for the breakup instabilities to fully develop before the bridge has already been stretched to a large aspect ratio. Fluid inertia stabilizes the bridge far beyond the quasistatic limit.

The events immediately following the initial rupture are detailed in Fig. 10. The initial perforations in the mucus film, which occur at the thickness minima as previously noted, rapidly expand due to surface tension. The remaining fluid, initially concentrated in the thicker regions of the bridge, is pulled into thin ligaments that span the newly formed holes. They continue to thin under the combined influence of surface tension and aerodynamic shear, as shown in the evolution from $t^* = 0.2$ ms to $t^* = 0.4$ ms. They ultimately break up into smaller droplets leading to a fine aerosol cloud, a process that is not resolved by the current model.

D. Effect of transglottal pressure

To investigate the influence of phonation intensity on the breakup process, a parameter study was performed by varying the transglottal pressure difference. Figure 11 presents the temporal evolution

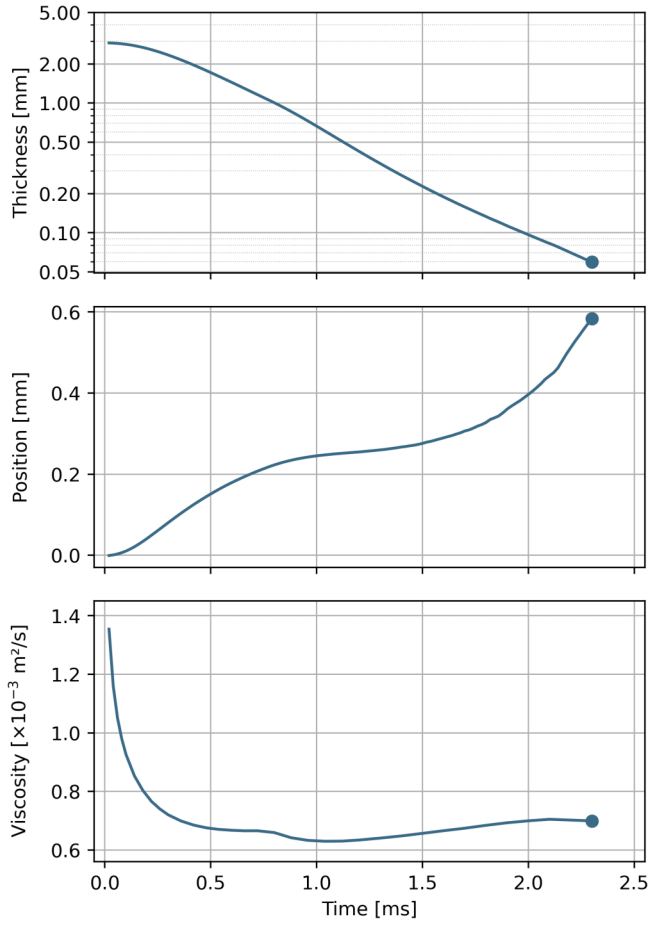


FIG. 8. Average mucus-bridge thickness (top), downstream position relative to the glottis (middle), and average kinematic viscosity (bottom) during the formation and rupture process.

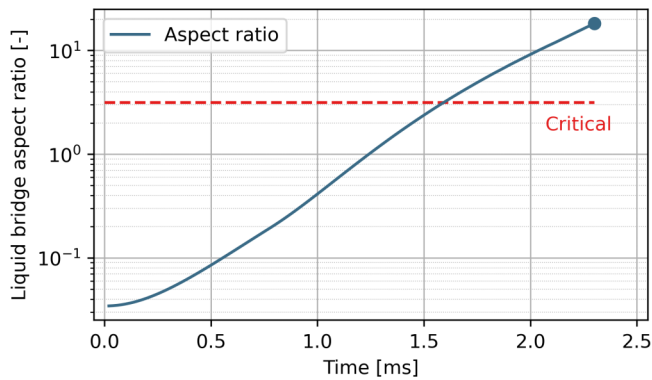


FIG. 9. Temporal evolution of the mucus-bridge aspect ratio (AR). The bridge ruptures at $AR \approx 20$, a value significantly exceeding the quasistatic Rayleigh-Plateau stability limit of $AR = \pi$ (red dashed line).

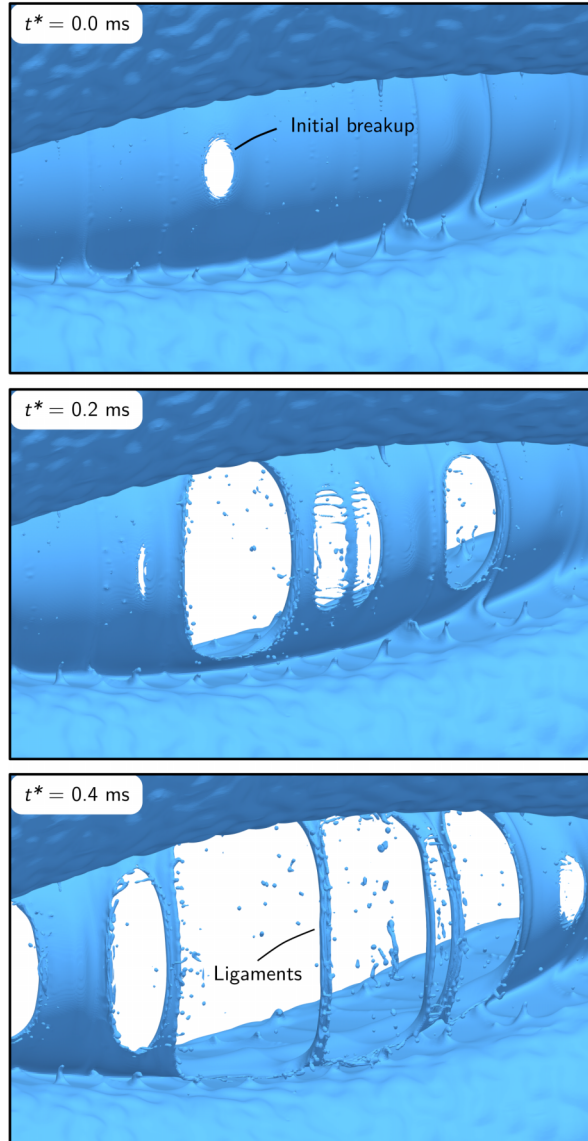


FIG. 10. Evolution of the mucus film immediately following initial breakup ($t^* = 0$ ms). Initial perforations in the film rapidly expand due to surface tension. The surrounding fluid is drawn into thin ligaments that span the newly formed holes at $t^* = 0.2$ ms. These ligaments continue to thin under aerodynamic and surface-tension forces at $t^* = 0.4$ ms, eventually leading to secondary breakup into smaller droplets.

of the mucus bridge's average thickness (top) and its downstream position relative to the glottis (bottom) for three pressure levels corresponding to Scherer *et al.* [11] (491 Pa, 981 Pa, and 1472 Pa, respectively).

As shown previously, the initial thinning of the mucus bridge, from its contraction up to $t \approx 2.0$ ms, is nearly independent of the applied pressure difference. During this phase, the thinning process appears to be dominated by the prescribed vocal-fold kinematics and the cohesive surface-tension forces. However, after $t \approx 2.0$ ms, when the bridge has thinned to a critical thickness of approximately 0.1 mm, the influence of aerodynamic forces becomes dominant and the behavior

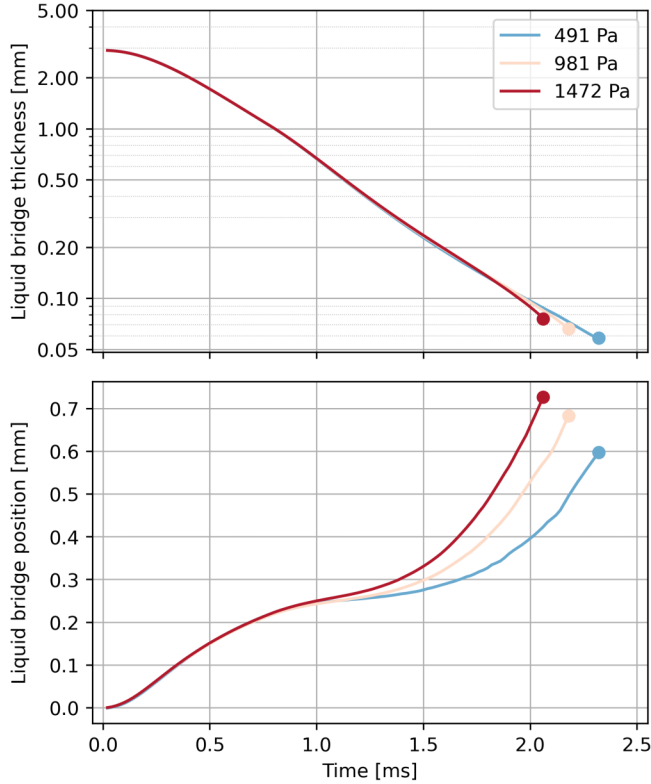


FIG. 11. Effect of transglottal pressure on mucus-bridge dynamics; data plotting ends at the moment of bridge breakup. (Top) The temporal evolution of the average bridge thickness shows that while initial thinning is pressure independent, higher pressures lead to earlier final rupture. (Bottom) The downstream position of the mucus bridge relative to the glottis shows that higher pressure results in a greater convection of the mucus bridge due to increased aerodynamic drag.

diverges significantly. For the highest pressure (1472 Pa), the strong aerodynamic forces lead to a rapid rupture at $t \approx 2.04$ ms. In contrast, at the baseline pressure of 491 Pa, the weaker aerodynamic forces take significantly longer to overcome the remaining fluid inertia and surface tension, delaying the final rupture until $t \approx 2.32$ ms. This indicates the existence of a critical thickness below which the bridge becomes highly susceptible to aerodynamic forces, and the magnitude of these forces then dictates the timing of the final breakup.

While the initial thinning rate is largely independent of pressure, the final breakup timing and bridge position are not. Figure 11 (bottom) shows that while the bridge is pushed downstream in all cases, the downstream velocity (represented by the slope of the curve) is substantially greater at higher pressures and the increased pressure load between upstream and downstream side of the mucus bridge. As a result, the initial breakup occurs earlier but further downstream.

IV. CONCLUSION

A numerical model was presented for simulating the formation and breakup of the mucus bridges between opening vocal folds. The three-dimensional, incompressible CFD model includes various physical effects such as surface tension, non-Newtonian mucus properties, and prescribed motion of the simplified vocal folds.

The numerical results show that introducing small-scale perturbations to the initial mucus film was necessary to trigger realistic and complex breakup of mucus bridges. Breakup was seeded at initial thickness minima, demonstrating a direct link between the initial surface topology and the final breakup structure. Analysis of bridge formation and breakup revealed exponential thickness decay, with initial rupture occurring at an aspect ratio of approximately 20. The downstream position of the mucus bridge is mainly dictated by (i) the prescribed opening of the vocal folds and (ii) the transglottal pressure difference. Furthermore, once the mucus bridge had contracted at the beginning of the process, non-Newtonian behavior showed only minor influence, as the average kinematic viscosity in the mucus bridge remained nearly constant. A parametric study showed that higher transglottal pressures, corresponding to increased phonation intensity, accelerated the final thinning and rupture of the bridge.

While the current approach successfully captures mucus-bridge formation and breakup, future work should consider specialized methods for thin-film rupture to ensure robust and physically consistent results across different flow conditions. Additionally, advanced meshing strategies with adaptive refinement could better resolve secondary-breakup cascade into the smallest droplets. Subsequent studies could also utilize fluid-structure interaction to model self-sustained oscillations and investigate effects of varying mucus rheology.

ACKNOWLEDGMENTS

The authors are grateful to the Center for Information Services and High Performance Computing (ZIH) at TU Dresden for providing its facilities for high-throughput calculations.

DATA AVAILABILITY

The data that support the findings of this article are openly available [54].

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