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A Fast Robust 1D Flow Model for a Self-Oscillating Coupled 2D FEM Vocal Fold Simulation

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Abstract

A balance between the simplicity and speed of lumped-element vocal fold models and the completeness and complexity of continuum-models is required to achieve fast high-quality articulatory speech synthesis. We develop and implement a novel self-oscillating vocal-fold model, composed of a 1D unsteady fluid model loosely coupled with a 2D FEM structural model. The flow model is capable of robustly handling irregular geometries, different boundary conditions, closure of the glottis and unsteady flow states. A method for a fast decoupled solution of the flow equations that does not require the computation of the Jacobian is provided. The model is coupled with a 2D real-time finite-difference wave-solver for simulating vocal tract acoustics and a 1D wave-reflection analog representation of the trachea. The simulation results are shown to agree with existing data in literature, and give realistic pressure-velocity distributions, glottal width and glottal flow values. In addition, the model is more than an order of magnitude faster to run than comparable 2D Navier-Stokes fluid solvers, while better capturing transitional flow than simple Bernoulli-based flow models. The vocal fold model provides an alternative to simple lumped-element models for faster higher-quality articulatory speech synthesis.

Index Terms: vocal fold model, finite element method, 1D flow solver, articulatory speech synthesis

1. Introduction

With the increase in simulation and computational capabilities, articulatory speech synthesis has once again become an active area of research. Articulatory speech synthesizers have used vocal tract models ranging from simple 1D analogues [1][2] to precise but computationally expensive simulations of pressure propagation in 3D vocal tracts [3]. Due to the computational cost involved, only 1D vocal tract models have been run at real-time simulation rates using adequate spatial resolution [4]. Recently Zappi et al [5], implemented a GPU-based 2D Finite-Difference Time-Domain (FDTD) solver to achieve real-time 2D speech synthesis. A critical component that all articulatory synthesizers require is a vocal fold model that acts as the excitation source to the acoustic simulation of the vocal tract. Most articulatory speech synthesizers have generally relied on parametric glottal models [6] or low-dimensional lumped-element models [7][8][9] of the vocal folds as the glottal source, as they are faster to compute.

Though early models were based on the *source-filter theory* [10], it is now well-established that the sound-source from the glottis and filtering by the vocal-tract articulators are nonlinearly coupled [11]. Thus, improved glottal models are critical to achieve high quality articulatory speech synthesis. Many

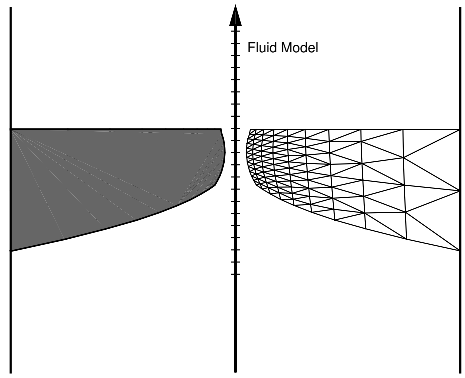

Figure 1: *Vocal Folds with Fluid Model along midline. Midline symmetry is assumed and only the right vocal fold is simulated*

types of vocal fold models have been implemented with different combinations of structural and flow models [12]. In particular, continuum models of the vocal folds can better capture the interaction between the vocal-fold structure and airflow [13]. A finite-element solver (FEM) is generally used for the structural mechanics, coupled with fluid dynamics solved using either the FEM or finite-volume method (FVM). For example, an early model developed by Alipour et al [14], was a quasi-3D continuum model where the FEM equations were solved in multiple 2D planes driven by a Bernoulli-based flow model. Later models and implementations included: a version driven by the 2D Navier-Stokes equations [15], quasi-3D strongly coupled fluid-solid simulations with collision handling [16], fully resolved flow computation [17] to extremely detailed but heavy full-3D models [18]. These quasi-3D and 3D models have shown themselves capable of reproducing many of the features that are seen in human phonation; however, the computational cost involved in solving both the solid and fluid equations in 2D and higher dimensions continues to be prohibitive. This is especially the case when simulating the complete articulatory speech synthesis pipeline. Previous comparisons of simplified Bernoulli-based solvers vs Navier-Stokes solvers for lumped-element [19] and continuum-models [20], showed that, while Bernoulli models do well in predicting bulk intraglottal pressures, they are unsatisfactory in computing the location of separation pressure and glottal flow rate.

There is the need for a balance between simple, fast lumped-element models and more complex continuum models. In particular there is a need for a fluid model that is computa-

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Table 1: *Model Parameters (Order: Body-Cover-Ligament)*

Parameter Name	Value(s)
Longitudinal Young's modulus	20 - 15 - 30 kPa
Transverse Young's modulus	2 - 1.5- 3 kPa
Longitudinal Shear modulus	12 - 11 - 20 kPa
Viscosity	6 - 3 - 5 poise
Longitudinal Poisson's ratio	0.4
Transverse Poisson's ratio	0.9
Lung pressure	1.0 kPa
Fluid density	1.14 kg/m^3
Fluid dynamic viscosity	1.86e-5 $Pa \cdot s$
χ_{min}	0.2

tionally cheaper than the solution of the 2D Navier-Stokes equation but can capture unsteady flow effects better than Bernoulli-based models. We propose a 2D FEM-based model that uses linear-elastic solid mechanics coupled with a 1D unsteady fluid model for modelling laryngeal flow. We aim to achieve similar performance to 2D Navier-Stokes solvers at lower computational costs. The model is coupled with a wave-reflection analog based trachea model and the afore-mentioned 2D FDTD vocal tract solver. The fluid model is robust, and stable coupled self-oscillation is achieved. The model can handle either pressure-based or velocity-based boundary conditions, and we present a method for a fast decoupled solution of the fluid equations. The model also fits in nicely with the 2D FDTD vocal-tract solver [5], enabling us to work towards building a complete articulatory speech-synthesizer. The paper is structured as follows: in section 2, we present the model formulation and implementation, followed by the simulation results in section 3. Finally, conclusion and possible future work in given in section 4.

2. Model Formulation

2.1. Solid Mechanics

We use a 2D structural model based on the model by Alipour et al [14] with a focus on lower computational load. We choose to assume symmetry across the midline plane and simulate a hemi-laryngeal model instead of both vocal folds. The model uses a linear-elasticity assumption which is computationally cheaper to solve, and has been validated in previous studies. A linear shape function is used for the finite-element formulation and the material properties are taken from a recent updated version of the model [21] (Table 1). The vocal fold mesh was the same mesh used by [14] and the vocal fold structure was divided into three material regions: body, cover and ligament. The FEM solution of the spatial problem yields a second-order matrix differential equation (1) in the time domain:

$$M\ddot{\Psi} + D\ddot{\Psi} + K\Psi = F \tag{1}$$

The equation is discretized using the second-order central scheme centred at the nth time-step for stability. The aerodynamic force and string forces are used to calculate forcing vector *F*, in equation (1). Since contact between the symmetric vocal folds would happen at the glottal midline, a rigid plane is assumed to be present there. When the vocal fold reaches the midline, an impact force is applied to prevent interpenetration. The contact force is normalized over a collision region defined by the A_{closed} value associated with the fluid model, that is explained in subsection 2.2.2.

2.2. Fluid Mechanics

The modelling of the flow through the vocal folds is a challenging problem for many reasons. The flow transitions from a mainly laminar region before the minimum glottal area, then separating into a jet regime after a flow-separation point. Another challenge is the constant closing and reopening of the glottal channel; this is a particular issue for volume-based 2D fluid models using the FEM or FVM, as the onus falls on the solid model to ensure there are no zero/negative volume fluid elements. In particular, we aim to find a 1D model that can handle these issues at a lower computational cost than a standard 2D fluid solver. We choose to modify the 1D fluid model proposed by Anderson et al [22], based on the ideas of Cancelli and Pedley [23]. The model was implemented for fluid-structure interaction simulations in the upper airway, for the purpose of simulating obstructive sleep apnea (OSA). The model is chosen as OSA simulations have many of the requirements that make glottal flow modelling difficult and gave similar results to 3D simulations at faster speeds.

The flow continuity and momentum equations for the model are written as:

$$\frac{\partial}{\partial t}A = \frac{\partial}{\partial x}Au = 0 \tag{2}$$

$$\rho u\frac{\partial}{\partial x}u + \rho\frac{\partial}{\partial t}u + \frac{\partial p}{\partial x} - \tau\frac{s}{A} = 0 \tag{3}$$

$$\tau - \tau_{fric} - \tau_{\chi} = 0 \tag{4}$$

where *s* is the perimeter around the cross-section area *A*, *u* is average velocity, *p* is pressure and ρ is density. τ models the viscous losses with $\tau_{fric} = -2\mu(s/A)u$ and $\tau_{\chi} = (A/s)(1 - \chi)\rho u(\frac{\partial}{\partial x}u)$. In particular, the χ term (which is 1 before the flow separation point and χ_{min} after it) is important for modelling the flow in vocal folds.

2.2.1. Numerical Implementation

In the case of a velocity-driven flow, we can see that the flow equations can be solved sequentially. The velocity boundary condition can be applied to 2, to calculate the velocity distribution $u(x)$. This can be used to compute $\tau(x)$ using 4. Finally, equation 3 is solved using the calculated u and τ values. This procedure is extremely fast, and computationally efficient. On the other hand, pressure-driven flows require a coupled-solution of equations 2-4, which can be solved using Newton's method. This method is more computationally intensive as it requires calculation of the Jacobian matrix and successive iterations. In addition, glottal flow is generally driven through pressure-pressure boundary conditions.

Thus, we suggest a method to convert pressure-pressure boundary conditions to the equivalent velocity-pressure boundary conditions, followed by a decoupled solve. The solution involves a bounded search, where we iterate the system till we find uInlet-pInlet boundary conditions equivalent to the specified pInlet-pOutlet boundary conditions. Our solution procedure is as follows:

1. Create two initial guesses for input velocity, $u1_{i=0}, u2_{i=0}$ for a given $pInlet^{n+1} - pOutlet^{n+1}$ boundary condition. The superscript refers to the simulation time, and the subscript refers to the iteration number. The previous time step's input velocity $uInlet^n$ and $1.1 * uInlet^n$ are good guesses to speed up convergence.

2. Use the decoupled solver to find the outlet pressures $p1_{i=0}, p2_{i=0}$ for the $u1_{i=0} - pInlet$ and $u2_{i=0} - pInlet$ systems respectively.
3. Calculate the change in pressure with velocity $dpdu_i = (p2 - p1)/(u2 - u1)$
4. Update $u1_i = u2_{i-1}$ and $p1_i = p2_{i-1}$ from the previous iteration, and create a new guess for $u2_i$ using $dpdu_i$.
5. Solve the decoupled equations for the $u2_i - pInlet$ boundary conditions
6. Calculate the difference between the target outlet pressure and the new computed outlet pressure as *diff*=$pOutlet - p2_i$
7. Iterate steps 3-6 until *diff* is below a certain tolerance value

2.2.2. *Fluid-Structure Coupling*

The structural and fluid models are coupled through the area-function that is the input to the fluid simulation, and the aerodynamic pressure that is used to compute the forcing vector for the FEM solid mechanics. We choose to loosely couple the solid and fluid models rather than have a combined formulation; this enables us to treat the area function as a pre-computed quantity for solving the 1D fluid model sequentially. The structural model is discretized in the coronal plane, with the fluid model computed along the centre-line/midline in Figure 1. Velocity components perpendicular to the midline are considered to be zero. A fourth-order asymmetric scheme is used for spatial discretization of the fluid model. At every discrete point, a cross-sectional area is extracted from the structural model, which is the medial-lateral opening between the two vocal folds, multiplied by the distance in the dorsal-ventral plane. The triangular nature of the glottal opening is taken into account when computing the cross-sectional area *A* and the associated perimeter *s* for the fluid model. Another important case, is that of collisions during the self-oscillation process. At every time-step, the updated area function is extracted from the structural model, and passed to the fluid model. Since equation 4 has terms divided by *A*, we choose to handle the fluid model area through warping as suggested in [22] with a "safe" Area function:

$$A_{safe}(x,t) = A(x,t) + A_{closed} * w(A(x,t)) \quad (5)$$

The transition function is defined as:

$$w(A(x)) = \begin{cases} 0, & \text{for } A(x) > A_{small} \\ \frac{A(x)-A_{small}}{A_{closed}-A_{small}}, & \text{for} A_{closed} \leq A(x) \leq A_{small} \\ 1, & \text{for} A(x) < A_{closed} \end{cases} \quad (6)$$

where A_{closed} is the smallest numerically stable area that is empirically determined, and A_{small} is the area at which transition begins. Anderson et al [22] suggested a value of A_{small} = 2.5*A_{closed}. The solid model is allowed to collide with the mid-line (as shown in Figure 1) and a collision force is applied to the surface nodes. To ensure stability, the node are assumed to be in a state of collision, while the minimum glottal area is lower than A_{closed}. The pressure at the surface nodes of the FEM mesh are assumed to be equal to the concurrent computed pressures at the glottal mid-line. A linear interpolation is used to find the pressure's at each surface node of the FEM body. The Force Vector on the node is then computed as follows:

$$\boldsymbol{F}_{node} = p_{node} * A_{node} * \hat{\mathbf{n}} \quad (7)$$

where p_{node} is the pressure at the node, A_{node} is the effective nodal area shared between the elements the node belongs to, and $\hat{\mathbf{n}}$ is the unit normal vector to the nodal surface. Both the fluid and structural models are temporally discretized using a central scheme with same time-stepping.

2.3. Vocal Tract and Trachea Model

The acoustic simulation of the wave propagation through the vocal tract is done through a 2D-FDTD simulation (as seen in [5]) while the trachea is simulated through a wave-reflection 1D model [24]. The glottal flow output U_g of the vocal fold model is fed as the input to the vocal tract simulation. The supraglottal pressure P_{sg} and subglottal pressure P_{sub} are read from the vocal tract and trachea models respectively and act to couple the systems.

3. Simulation Results

We first tested the performance of our decoupled solution scheme vis-a-vis the full coupled-equations 1D solver; both systems were coded in MATLAB and used to characterize flow through a forced oscillating vocal fold model. On average our system was 45-50 times faster for a range of discretization values. The system was also compared to a finite-difference based 2D unsteady Navier-Stokes solver for flow through a uniform pipe with time-varying boundary conditions; this simpler comparison example was chosen due to the paucity of full-fledged 2D Navier-Stokes solvers in MATLAB. Our model was again more than an order of magnitude faster than the 2D Navier-Stokes solution.

The fluid model was then used to self-oscillate a coupled 2D structural vocal fold model. Of particular interest is how realistic are the pressure-velocity predictions of the 1D model. We compare our model to data published by Alipour et al [14] [21], where an established linear-elastic structural model was driven by an unsteady 2D Navier-Stokes flow model. While the latter model contains the false vocal-folds as well, we expect the results in the laryngeal region to be qualitatively and quantitatively similar. The vocal fold simulations are run at the audio rate of 44.1 kHz, coupled to both the vocal-tract and trachea models. Tubes representing the vocal tract shapes for the vowel /a/ were created based on the area function data extracted from magnetic resonance imaging data by Story et al [25]. The 1D fluid model took an average of 3-4 iterations per time-step to first convert the boundary conditions to a velocity-driven flow and obtain the final solution. The lack of sudden variations in the boundary conditions when simulating at small time-steps helped convergence occur quickly.

The fundamental frequency of the model was approximately 146 Hz, with a peak flow rate U_g of 473 mL/s, mean flow rate of 241 mL/s and maximum glottal width of 1.67 mm. To better understand if the fluid model is predicting reasonable pressure-velocity distributions in time and space during the phonation cycle, we focus on output values at specific phonation frames. Figure 2 shows characteristic vocal fold shapes at different time steps during the self-oscillation process; Figures 3 and 4 plot the corresponding spatial pressure and velocity distributions. The three shapes chosen, correspond to a convergent glottis shape, divergent glottis shape and an intermediate shape respectively.

We can see that our model produces realistic pressure and

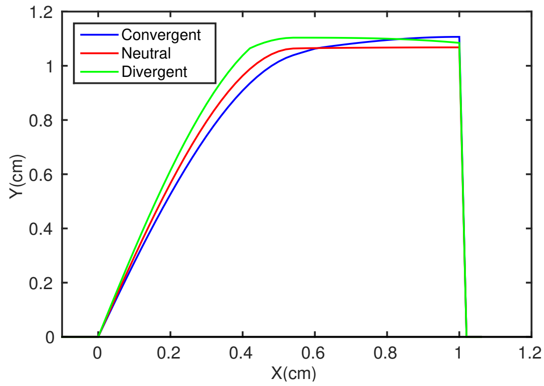

Figure 2: *Vocal fold shapes during phonation cycle. 1:Convergent Glottis, 2:Divergent Glottis, 3:Intermediate Glottal Shape*

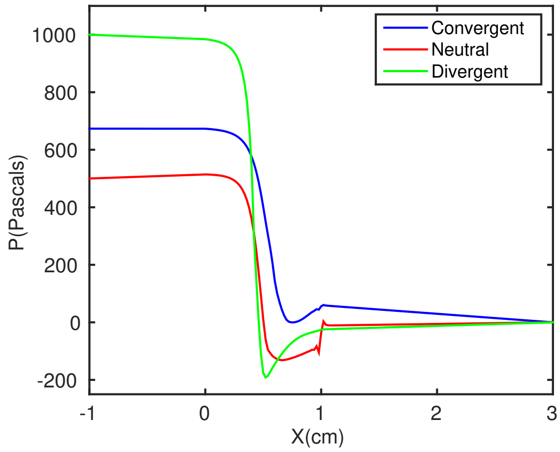

Figure 3: *Midline pressures from fluid model*

velocity distributions. The peak velocities correspond closely to the smallest glottal diameter, and decrease sharply from a higher peak in the divergent glottis, as expected. We see that our velocities have a sharp tail-off to a jet regime at the outlet of the vocal fold due to the χ transition function; this is different to the values of Alipour et al [21] since they also have false vocal-folds and more accurately model turbulences in their 2D model. However, our pressure distributions show excellent convergence to the paper values. The divergent glottis shows a strong dip in pressure near the minimum constriction area with a realistic pressure recovery, while the convergent glottis have high pressures withing the glottis and decreasing pressures towards the end. There is a sudden pressure spike at the exist of the vocal fold in the neutral vocal fold shape; this is most likely a numerical imprecision arising from using a χ transition function with an internal step function. Table 2 shows the similarity between the peak velocity and pressure values predicted by our model and those seen in Alipour et al's [21] paper.

4. Discussion and Conclusion

In this paper, we have presented the formulation for a novel 1D fluid model that is used to drive a 2D continuum vocal fold model and preliminary simulation results. A linear-elastic assumption is used for its material properties and the finite-element method is used solve the solid mechanics problem. The fluid model, initially designed for flow in collapsible tubes, pre-

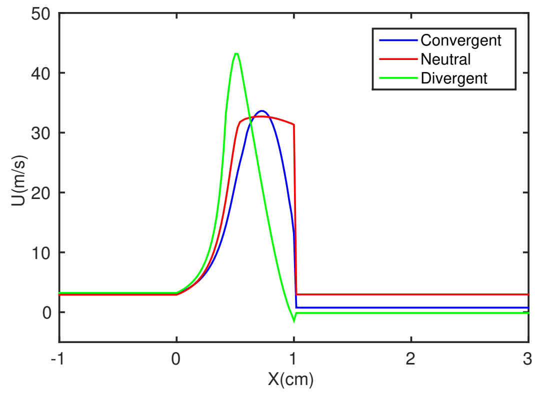

Figure 4: *Midline velocities from fluid model*

dicts the aerodynamic pressure that provides the forcing vector for the FEM solver. The solid and fluid mechanics are loosely coupled, enabling us to treat $A(x, t)$ as a known quantity for the flow simulation. The fluid model uses a fourth-order asymmetric spatial scheme and both models are temporally discretized using a second-order central scheme for stability. An area-warping model is implemented to prevent numerical instabilities; this makes the 1D fluid model well-suited to complete closure and reopening of complex uneven geometries seen in the glottal region, and remains stable during collisions.

Table 2: *A comparison of peak centerline velocities and subglottal pressures of the model presented in the paper (V_p, Ps_p) and from Alipour et al [21] (V_a, Ps_a). Note that the values from literature are estimates from published graph data*

Type	**Max V_p**	**Max V_a**	**Max Ps_p**	**Max Ps_a**
Convergent	34 m/s	33 m/s	670 Pa	690 Pa
Neutral	33 m/s	32 m/s	520 Pa	515 Pa
Divergent	43 m/s	45 m/s	990 Pa	950 Pa

A fast solution strategy to convert pressure-driven boundary conditions to velocity-pressure boundary conditions is provided. This allows us to solve the computationally cheaper velocity-driven de-coupled form of the nonlinear system of partial differential equations for the fluid model. The χ term enables us to approximate the location of pressure losses and flow separation, which steady Bernoulli-based flow models struggle with. The vocal fold model is coupled to a 2D FDTD vocal tract solver for the vowel shape /a/ and a 1D trachea wave-reflection analog. The model is computationally cheaper than 2D models and predicts pressure and velocity distributions that are similar to literature data from 2D Navier-Stokes simulations. The model provides promising initial results at a lower computational cost; however, further tests looking at the glottal flow U_g, phonation onset frequency, and sound quality of the overall system are needed to validate the model. The current implementation may also be optimized to target further performance improvements. Potential options include methods to speed up the matrix solver along with parallelization of the fluid model and the trachea model.

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