Experimental Validation of Direct and Inverse Glottal Flow Models for Unsteady Flow Conditions

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Abstract

The pressure drop along the glottal constriction drives vocal folds self-sustained oscillations during phonation. Physical modeling of phonation is classically assessed with the glottal geometry and the subglottal pressure as known input parameters. Several studies including in-vitro validation show that simplified one-dimensional flow models allow predictions of the flow characteristics to a fair extent. Application of physical modeling to study phonation abnormalities and pathologies requires input parameters which can be related to in-vivo measurable quantities commonly corresponding to the physical model output parameters. The current paper considers the inversion of some popular simplified flow models in order to estimate the subglottal pressure, the glottic constriction area or the flow model parameters under unsteady flow conditions. The theoretical predictions are tested against in-vitro measurements.

1. Introduction

Physical modelling of phonation or voiced sound production intends to predict the vocal folds oscillatory behavior in terms of relevant physical and physiological quantities. Classical physical phonation models estimate the vocal folds vibrations from simplified lumped models with a large number of parameters. Since the two mass vocal fold model several improvements are proposed in order to approach the complex reality of human voice production. Improvements have been realized among others concerning the mechanical description of the vocal folds \cite{1}, the glottal flow model during phonation \cite{2} or the acoustical coupling of the vocal folds to the vocal tract \cite{3}. In general, more advanced representations of the glottal cycle during phonation necessitate the development of more complex models with an increasing number of parameters and assumptions \cite{4}. Moreover, physical speech production models need control parameters which can be linked to human motor control \cite{5} in order to reproduce different types of voices. Furthermore, physical modeling in particular interesting for the study of irregular phonation patterns \cite{6}, effects of vocal fold asymmetries \cite{7} and vocal folds dysfunction due to pathologies \cite{8}. The influence of the applied parameter set on the model outcome is often assessed following an analysis-by-synthesis approach \cite{9} and further compared to in-vivo measured quantities. Despite the incessant development of clinical in-vivo measurement techniques, the observation of phonation in either normal or disordered \cite{10} conditions remains a difficult task often depending on invasive measurements methods. Consequently, the subglottal pressure, which is the main input parameter for standard physical phonation models, is difficult to be obtained directly from in-vivo measurements \cite{11}. Therefore the interest of inverting classical physical phonation models is multiple. Apart from a purely scientific interest, non-invasive in-vivo measured data with an unambiguous physiological or clinical meaning can be directly applied as a model input. At long term, inverse physical models might be validated on in-vivo data and eventually be applied in pathological conditions or for advanced voice synthesis. Moreover from a clinical perspective or for voice re-education purposes inverse models allow to account for subject-dependent data and hence for inter-subjects differences. Most of the studies considering inversion of phonation models retrieved in literature deal with inverse filtering techniques \cite{12}. Considering the inversion of bio-mechanical models implying both a mechanical and a flow part, an encouraging result is presented in \cite{13} where tissue characteristics are deduced from flow measurements. The current paper deals with the inversion of simplified one-dimensional flow models \cite{14} widely applied in classical direct phonation models \cite{2,3}. The aimed models are derived from Bernoulli’s one-dimensional flow equation corrected for viscous effects and flow unsteadiness. Application of the simplified flow theories in order to estimate important phonation quantities as phonation onset threshold pressure and oscillation frequency is positively validated on in-vitro glottal replicas \cite{16}. At first direct and inverse flow models presented in \cite{17} are summarized. Next the inverse models are validated on data obtained with moving rigid glottal replicas for different glottal constriction shapes.

2. Glottal flow models

2.1. One-dimensional model

Under the assumptions of one-dimensional, laminar, inviscid and incompressible flow, the Bernoulli’s equation gives the pressure difference $p_a - p_b$ between two points $a$ and $b$ of a constriction as following:

$$\Delta p_D(a,b) = \frac{1}{2} \rho \Phi^2 \frac{h_2^2 - h_1^2}{h_2^2}$$

\begin{equation}
(1)
\end{equation}

where $\Phi$ is the volume velocity (assumed constant along the constriction), $l_c$, the width of the constriction, $h_a$ and $h_b$, the heights of the constriction at the points $a$ and $b$, and $\rho$, the fluid density.

2.2. Viscosity correction

Due to the fluid viscosity, it has been shown \cite{2} that the flow separates from the walls and forms a jet in the diverging down stream part of the constriction at a position corresponding to the height $h_s = c_s \times h_b$, with $h_b$, the minimal height of the constriction and $c_s$, an ad-hoc separation coefficient greater than
1. Consequently, (1) only holds down to the separation point and the pressure in the constrictions after this separation point is considered equal to the downstream pressure. Moreover, for small constriction heights, viscous effects cannot be neglected. The pressure drop induced by viscous rubbings along the walls is expressed as [18]:

\[ \Delta p(a, b) = \frac{\mu}{L_g} \int_a^b \frac{dh(x)}{dx} dx \]  

where \( \mu \) is the dynamic viscosity of the fluid and \( a < b \). The Poiseuille term \( \Delta p_P(a, b) \) assumes a non-uniform two-dimensional velocity profile and therefore presents a viscosity related correction to the one-dimensionality assumed in (1).

2.3. Unsteadiness

Although several in-vitro experimental studies confirm the quasi-steady approximation made in (1) [19], pressure differences induced by flow unsteadiness due to wall movements \( h(x, t) \) or volume airflow variations \( \Phi(t) \) in time might be important at high frequencies. The corresponding pressure loss is expressed as (3):

\[ \Delta p_U(a, b) = \frac{\rho}{L_g} \int_a^b \frac{d}{dx} \left( \frac{\Phi(t)}{h(x, t)} \right) dx \]  

2.4. Direct model input, output and parameters

In classical direct flow models the upstream pressure \( p_0 \) is the main driving control parameter and therefore supposed to be a known input variable. The downstream pressure \( p_1 \) is assumed to equal atmospheric pressure, i.e. \( p_1 = p_a \). Apart from the mentioned pressures, the constrictions geometry schematically illustrated in Fig. 1 is assumed to be known. The model outcome is the one-dimensional pressure distribution \( p(x) \) along the geometry and the volume velocity whereas the separation coefficient and the required physical constants are known model parameters. The pressure \( p_g \) at the minimum constriction height \( h_g \) is predicted from (4) where the Poiseuille and unsteadiness terms are the previously defined optional corrections:

\[ p_g = p_0 - \Delta p_B(0, g) - \Delta p_P(0, g) - \Delta p_U(0, g) \]  

3. Inversion of the flow models

3.1. Inverse model input, output and parameters

In the searched inverse flow models, the role of input/control and output parameters defined for the direct model as respectively the upstream pressure and the pressure distribution within the constriction, and in particular \( p_g \), are interchanged. Therefore in the inverse flow model, \( p_g \) is a known input/control quantity. From the direct flow models previously described, three different cases of inverse problems are defined with respect to physical phonation modeling. Firstly, the upstream pressure \( p_0 \) is the unknown quantity, secondly, the minimum constriction height \( h_g \) is searched and finally, the parameter \( c_s \) is estimated from known \( p_0, p_g \) and \( h_g \).

3.2. Inversion methods

Depending on the case, different inversion strategies are applied. In case of the Bernoulli model, it can be seen from (1) that inversion is easily obtained analytically. When the additional terms (2) and (3) are added to the Bernoulli flow description, numerical iterative methods [20] are applied in order to invert the models. The model inversion becomes a classical minimization problem which can be solved using e.g. Newton’s algorithm. By considering \( p_g \) as a function of \( p_0, h_g \) and \( c_s \), i.e. \( p_g = f(p_0, h_g, c_s) \) the estimation of the inverted separation coefficient \( c_s^{inv} \), for instance, from known \( p_0, p_g \) and \( h_g \) values is given by:

\[ \|p_g - f(p_0, h_g, c_s^{inv})\| < \epsilon \]  

where \( \epsilon \) is the tolerance of the convergence process. For steady models, the inversion process can be applied independently on each value of a given signal over time. For unsteady models, as equation (3), the inversion process must be applied on the entire signal in order to avoid an error propagation on the inverted values, since the predictions at instant \( t = i \) are dependent on the predictions at previous instants \( t < i \).

4. Experimental set-up

In order to test the validity of the presented direct and inverse models, an experimental set-up including a moving rigid constriction replica was used [21]. Two different geometrical constriction shapes were considered: uniform (with round entrance) and round as depicted in Fig. 2. Steady flow was provided by a valve controlled air supply. For each shape, unsteady flow is generated by the driven movement of one of the rigid vocal fold replicas, for frequencies in the range 3-30Hz and 0.6mm amplitude, corresponding to a Strouhal number, Sr, in the range \( 10^{-3} - 10^{-2} \). Pressure transducers (Endevco, Kulite) positioned in pressure taps upstream of the replica and at the minimum constriction height allowed to measure \( p_0(t) \) and \( p_g(t) \). An optical sensor was used to measure the constriction height \( h_g(t) \).
5. In-vitro measurements and discussions

5.1. Uniform constriction

A uniform constriction shape is particularly interesting to evaluate the Poiseuille correction without interference of the applied separation coefficient \( c_s \), since flow separation occurs at the constriction end. For small constriction apertures, the Poiseuille correction becomes important and allows to predict a positive pressure at the minimum constriction height, whereas the steady Bernoulli model predicts no pressure drop regardless the upstream pressure. As shown in Fig. 3, the predictions made by the direct models including the Poiseuille term are close, with a mean error less than 20%, to the experimental data. These good direct models performances enable to retrieve the minimum constriction height \( h_g \), with a mean error less than 15%, from the measured \( p_0 \) with the inverse models. The small errors made on direct predictions lead to more important errors on the inverted values of the upstream pressure \( p_0 \), with a mean error more than 40%. On the other hand, neither steady and unsteady Bernoulli direct models are able to predict the measured pressure \( p_g \) in a suitable way so that the inversion can not be considered (and omitted from Fig. 3). It can also be observed that the addition of the unsteadiness term in the model does not quantitatively affects the yielded predictions, although the maximum \( p_0(t) \) values are shifted in accordance with the measured values.

![Figure 3: Measurements and models predictions for the uniform shaped vocal folds vibrating at 25 Hz (Sr ≈ 10^{-2}) : (top) minimum constriction height \( h_g \), (middle) pressure at the minimum constriction height \( p_g \) and (bottom) upstream pressure \( p_0 \).](image)

5.2. Round constriction

As well as for the uniform constriction geometry, the direct model performance increases when accounting for viscosity. This can be seen comparing the predicted and measured pressure presented in Fig. 4 and is in particular crucial in estimating the timing as well as the maximum values of \( p_0 \). For geometries involving a diverging part, the choice of the separation coefficient \( c_s \) determines the prediction accuracy. As shown in Fig. 4, the direct model outcome with \( c_s = 1.2 \), which is frequently used to predict the separation point, overestimates the measured pressure drop at the minimum constriction height. The inverted estimation of \( c_s \), illustrated at the bottom of Fig. 4, from the measured pressure \( p_g \) appears to be around 1.08 for the Bernoulli+Poiseuille models whereas the estimation of \( c_s \) from the Bernoulli models does not give a consistent result, which is expected considering the poor direct models predictions.

![Figure 4: Measurements and models predictions for the round shaped vocal folds vibrating at 25 Hz (Sr ≈ 10^{-2}) : (top) minimum constriction height \( h_g \), (middle) pressure at the minimum constriction height \( p_g \) (computed with \( c_s = 1.2 \)) and (bottom) inverted separation coefficient \( c_s \).](image)

Using the separation coefficient \( c_s = 1.08 \) and including the viscosity related correction in the direct model allow to predict precisely, with a mean error less than 5%, the pressure at the minimum constriction height \( p_g \) as shown in Fig. 5. Besides, the corresponding inverse models give accurate predictions, with also a mean error less than 5%, of the upstream pressure from the measured pressure \( p_0 \). But it also appears that the inverted minimum constriction height can be widely overestimated due to a small difference between the predicted and the measured values of \( p_0 \). In this case, the inversion process leads to an error amplification, with a mean error on the inverted \( h_g \) about 20%.

6. Conclusions

The inversion of simplified flow models applied in well known physical vocal folds models is formulated and the model performance is tested towards moving in-vitro replicas of typical vocal folds shapes mounted in a suitable experimental set-up. The influence of viscous effects and unsteady flow effects due to wall vibrations on the prediction of the pressure at the minimum constriction height is considered. Thus the addition of a viscosity related correction in the one-dimensional flow model allows a major improvement of the predictions accuracy compared to in-vitro measurements, which reveals the weakness of the simple Bernoulli model. On the other hand, including the unsteadiness term does not visibly affect the results of the direct flow models. The correctness of the inverse models is strongly dependent on the accuracy of the direct flow models. Therefore, concerning one-dimensional simplified flow models based...
on the Bernoulli’s equation, the choice of the ad-hoc separation coefficient appears to have a significant influence on the accuracy of the predictions. Accordingly, it seems interesting to continue the flow models inversion by validating the value of the separation coefficient from more advanced glottal flow models, like two-dimensional, boundary layer or three-dimensional models.

7. Acknowledgements

This work has been supported by a PhD grant from the French Ministry of Education and Research. The authors also want to thank A. Hirschberg, Technical University of Eindhoven, The Netherlands, for kindly providing the moving rigid vocal folds replica and D. Blanc and D. Rey, Département Automatique, GIPSA-Lab, France, for their help in building the experimental set-up.

8. References