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EXPERIMENTAL MODELLING OF PHONATION USING ARTIFICIAL MODELS OF HUMAN VOCAL FOLDS AND VOCAL TRACTS

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Abstract: *The study provides information on experimental research on a complete 1:1 scaled model of human phonation. The model includes human lungs, the trachea, the laryngeal part with artificial vocal folds and the vocal tracts designed for different vowels. The measurement set up enables modelling the time signals not easily measured in humans during phonation as for example fluctuations of the subglottic, laryngeal and oral pressures measured simultaneously with the glottis opening and the glottis area registered by a high-speed camera. The simulation of phonation is performed in the ranges of the airflow rate and the subglottic pressure typical for a normal humans' physiology.*

Keywords: Biomechanics of voice, Experimental simulation of human phonation in vitro.

1. Introduction

The contribution presents the special experimental facility designed for in vitro measurements of voicing performed on originally developed 1:1 scaled models of human vocal folds and vocal tract. The designed models are based on CT and MRI measurements of human subject during phonation. The measured phonation (aerodynamic, vibration and acoustic) characteristics are comparable with values found in humans; however, obtaining reliably some of such characteristics, like for example pulsations of air pressure in subglottal, intraglottal and supraglottal spaces, from in vivo measurements is usually very problematic or nearly impossible. The knowledge of these characteristics for the artificial vocal folds can be useful for experimental verification of 3D computational finite element models of phonation due to relatively exactly defined input aerodynamic, material and geometrical parameters, which is also problematic to know reliably and exactly in humans.

2. Methods

Comparing to the recent studies of the authors (Horáček et al., 2016a, b) the measurement set up was updated, especially for measurements of air pressures in the laryngeal part of the vocal tract near the vocal folds and in the oral cavity, see Fig. 1. The high speed camera had to be included in the measurement set up for analyses of the vocal folds vibration. The developed simplified model of the human lungs, which includes the splitting of the air spaces up to the fourth order branching, was also built-in in the experimental facility, see Fig. 2. The measurements presented in this study were performed with an innovative three layer vocal folds model, see Fig. 3. Silicon wedge, modelling a vocal fold body (3), was added inside the vocal fold reducing the space of the liquid layer modelling the lamina propria layer (2) positioned under the silicon cover (1). This modification substantially reduced the airflow rate of phonation onset below $Q = 0.1$ l/s and the fundamental phonation frequency below $F_0 = 80$ Hz. The plexiglass model of the vocal tract for the vowel [u:], designed on the basis of CT images measured during a subject phonation is attached to the larynx model at the bottom end, see Fig. 4. Installation of pressure transducers in the vocal tract cavity allowed obtaining more information on magnitudes of pressure fluctuations in laryngeal and oral parts of the vocal tract. Usage of the high-speed camera enabled to study vibration amplitudes of the vocal folds and to determine the glottis opening and closing phases in relation to the acoustic and pressure data measured synchronously in time domain. The all

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measured signals were simultaneously sampled by the frequency of 16.4 kHz and registered by the measurement system Brüel & Kjaer PULSE controlled by a personal computer (see Fig. 1).

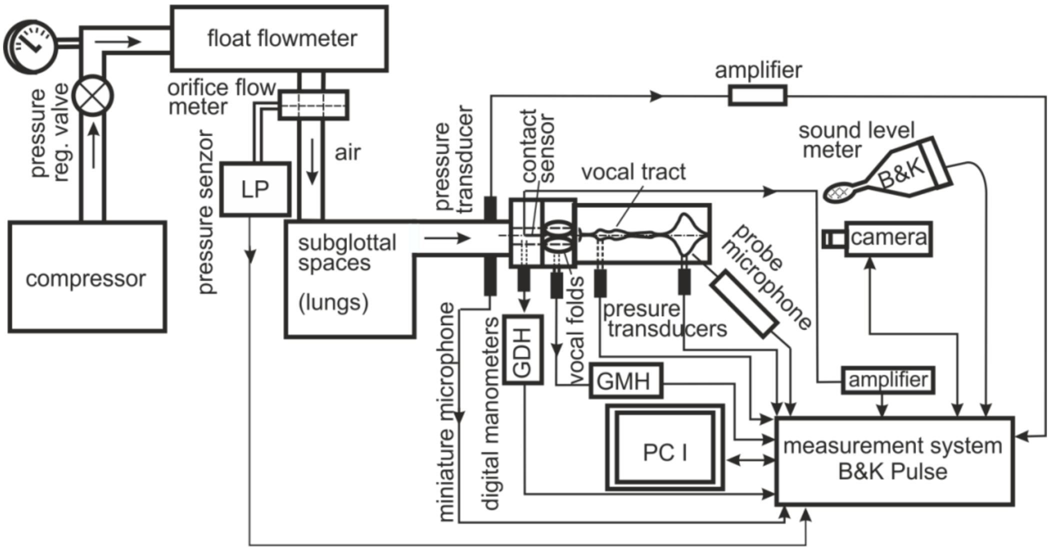

Fig. 1: Scheme of the measurement set up.

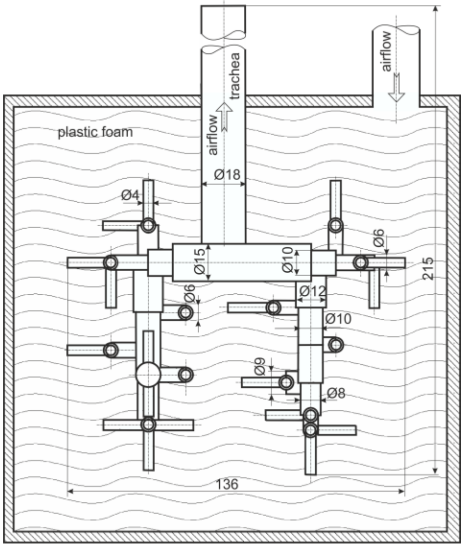

Fig. 2: Schema of the lungs model.

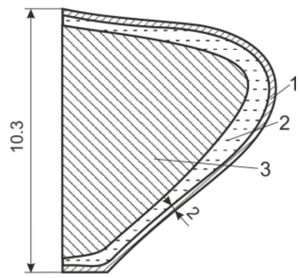

Fig. 3: Schema of the vocal fold model.

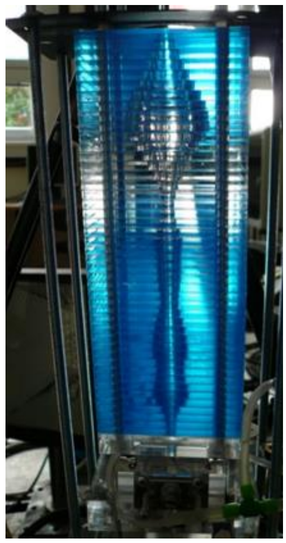

Fig. 4: Vocal tract model.

3. Some examples of results

The example of one measurement trial is demonstrated by Fig. 5. Each trial, which usually takes 10 s, is performed for a constant preset airflow rate. All measured time signals are monitored on the PC screen and the measurement starts when the phonation is stabilized. After about two seconds powerful lights, focused on the self-oscillating vocal folds, are switched on, which enables for about 0.5 s to register and later to visualize the vibrations of the vocal folds by the high speed camera. However, the light influences some pressure signals by a temperature drift due to an intensive heating of the pressure transducers positioned near the vocal folds. After about 2 s the lights are switched off which is followed by a sudden stop of the airflow, and the distorted signals are returning slowly back to zero or to a constant equilibrium value. The drifts of the pressure signals are corrected afterwards, during evaluation of the results.

Fig. 6 shows typical time records of the following measured signals: 1/ the subglottal pressure measured just below the vocal folds by two different pressure transducers for determination of the pressure fluctuations (the signal denoted by Mic_{Min}) and the mean pressure value (the signal denoted by P_{sub}), 2/ the laryngeal pressure (P_{larynx}) measured just above the vocal folds, 3/ the oral pressure (P_{oral}) measured in the mouth cavity of the vocal tract model, and 4/ the acoustic sound signal (*Mic*) measured outside the vocal tract of about 30 cm from the mouth. These signals are shown together with 5 images of the glottis taken during one period ($T_{period} = 1 / F_0$) of the vocal folds vibration. The glottis was closed for 41 % of the period followed by the maximum of the glottis opening *maxGO* = 0.8 mm.

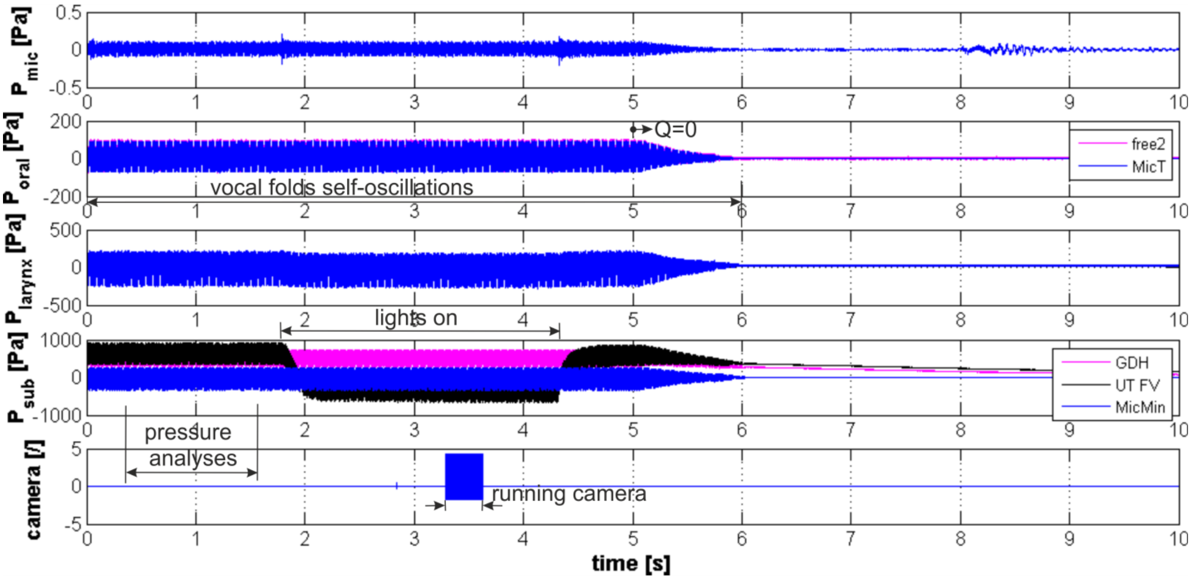

Fig. 5: Example of registered signals during one measurement procedure - airflow rate Q = 0.04 l/s.

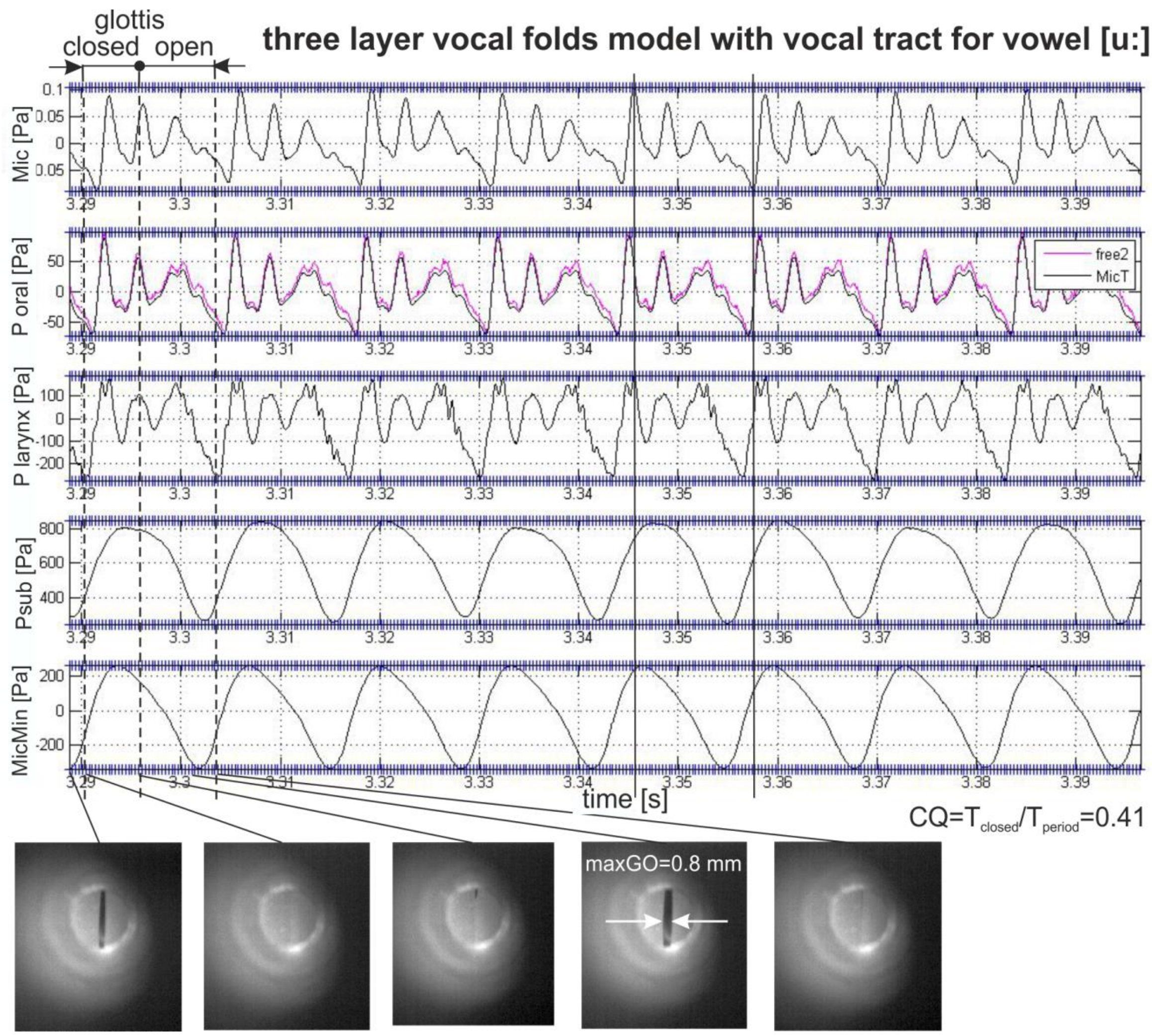

Fig. 6: Example of the measured pressure signals and snapshots of the vocal folds vibration (mean flow rate Q = 0.04 l/s, closed quotient CQ = 0.41, maximum glottis opening maxGO = 0.8 mm).

The maxima of peaks of the generated sound level (*Mic*) appear after the glottis closure. The pressure waveforms inside and outside the vocal tract (P_{larynx}, P_{oral}, *Mic*) are similar, only slightly time shifted due to the limited sound propagation velocity. These waveforms contain many higher harmonics of the fundamental phonation frequency and the resonance (formant) frequencies of the acoustic cavities of the vocal tract model for the vowel [u:], contrary to the signal (Mic_{Min}) for the subglottic pressure fluctuations where the fundamental frequency $F_0 = 76$ Hz principally dominates, see Fig. 6.

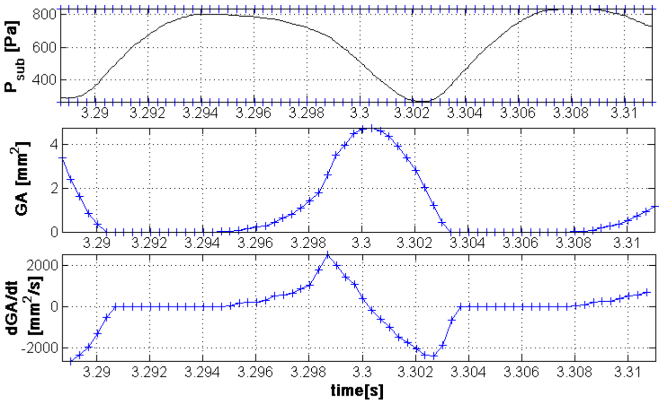

Fig. 7: Measured pressure signal and glottis area.

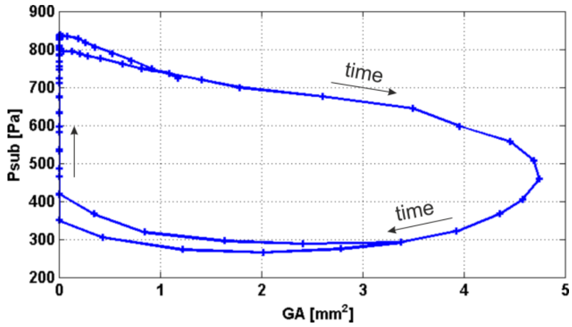

Fig. 8: Measured hysteresis loop.

Fig. 7 shows the evaluated glottis area *GA*(*t*) and the time derivative of the *GA* in relation to the measured subglottic pressure during of about one and half period of the vocal fold vibration. After the glottis closure for $t > 3.29$ s ($GA = 0$) the subglottic pressure increases and reaches the maximum value of about 800 Pa. Afterwards the glottis area increases during opening phase of the glottis and the subglottic pressure decreases to its mean value at the maximum opening of the glottis. During the closing phase of the glottis the subglottic pressure decreases to a minimum value. The maximum of the absolute value of the derivative *dGA*/*dt* appears before the glottis closure, i.e. shortly before the vocal folds collision. Interesting result is that this maximum negative value is practically identical with the positive maximum of the derivative *dGA*/*dt* observed in the opening phase. The derivative of the glottis area *dGA*/*dt* is an important parameter for estimating the impact stress between the colliding vocal folds.

Fig. 8 shows a nearly periodic dependence of the measured subglottic pressure on the glottis area. The time signals $P_{sub}(t)$ and *GA*(*t*) create a hysteresis loop oriented in time domain in the clockwise direction. The loop corresponds to the instability of the aeroelastic system due to the vocal folds self-oscillations; the energy from the airflow is transferred to the viscoelastic structure of the vocal folds.

4. Conclusions

Experimental modelling of phonation is important not only for understanding basic principles of voice production and for verification of mathematical models of phonation but also for detection of laryngeal pathologies and treatment of laryngeal disorders. The experimental modelling of phonation includes nearly all aspects of this complex physiological as well as physical process covering the fluid-structure-acoustic interaction phenomena.

Acknowledgement

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