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IMPACT STRESS IN WATER RESISTANCE VOICE THERAPY. A PHYSICAL MODELLING STUDY.

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Abstract:

Objectives: Phonation through a tube in water is used in voice therapy. This study investigates whether this exercise may increase mechanical loading on the vocal folds.

Study design: Experimental modelling study.

Methods: A model with 3 layer silicone vocal fold replica and a plexiglas vocal tract set for the articulation of vowel [u:] was used. Impact stress (*IS*) was measured in three conditions: For [u:] (1) without a tube, (2) with a silicon Lax Vox tube (35 cm in length, 1 cm in inner diameter) immersed 2 cm in water, and (3) with the tube 10 cm in water. Subglottic pressure and airflow ranges were selected to correspond to those reported in normal human phonation.

Results: Phonation threshold pressure was lower for phonation into water compared to [u:] without a tube. *IS* increased with the airflow rate. IS measured in the range of subglottic pressure, which corresponds to measurements in humans, was highest for vowel [u:] without a tube and lower with the tube in water.

Conclusions: Even though the model and humans cannot be directly compared, for instance due to differences in vocal tract wall properties, the results suggest that *IS* is not likely to increase harmfully in water resistance therapy. However, there may be other effects related to it, possibly causing symptoms of vocal fatigue (e.g. increased activity in the adductors or high amplitudes of oral pressure variation probably capable of increasing stress in the vocal fold). These need to be studied further, especially for cases where the water bubbling frequency is close to the acoustical – mechanical resonance and at the same time the fundamental phonation frequency is near to the first formant frequency of the system.

Key Words: phonation into a tube - vocal exercises - biomechanical loading - vocal fatigue; biomechanics of voice

Introduction

Phonation through a tube into water is a well-known voice therapy and training technique, especially in Scandinavia. The technique has gained popularity also in other countries during the last decade. The first papers about the technique appeared circa fifty years ago [1,2]. Afterwards many studies have been conducted describing the method itself [3-8], the effects of phonation through a tube in water on human subjects [9-21] or on models [22,23].

Water resistance therapy is typically performed by phonating through either a resonance tube made of glass, 26-28 cm in length, 9 mm in inner diameter [1,2] or a silicon so-called ‘Lax Vox tube’, length 35 cm, inner diameter 1-1.2 cm [5]. Resonance tube is recommended to immerse 2 cm in

water for the treatment of e.g. hyperfunctional voice disorders, while a deeper immersion, up to 10 cm or even 15 cm in water has been used to treat hypofunction, e.g. unilateral vocal fold paresis [8]. Lax Vox has been recommended to submerge 2-7 cm in water [5]. Phonation into a tube increases air flow resistance, the more so if the tube is long or especially if it is narrow, see e.g. [3,10]. It is well known that the depth of immersion of the tube in water regulates the airflow resistance, see e.g. [15, 21,23,24]. Increased airflow resistance increases intraglottal airpressure and thus tends to reduce collision between the vocal folds during phonation [3,4,6]. Modelling results and some electroglottographic (EGG) observations on humans support this, see e.g. [13,15,18]. However, some opposite results have also been obtained. According to the high-speed and EGG results by Laukkanen et al. [7] open time of the glottis decreased and contact quotient (*CQ*) increased for some subjects when they phonated into a long tube the distal end in air (60 cm or 100 cm in length, 2.5 cm in inner diameter). The EGG results by Tyrmi et al. [20] also showed that in some cases the contact quotient was higher in subjects phonating through a tube immersed 10 cm in water compared to normal vowel phonation, and in some cases it resembles the *CQ* found in loud phonation without a tube. Similarly, Guzman et al. [19] reported that in some subjects closed quotient and closing quotient increased when phonating into water through a silicon tube (45 cm in length, 2 cm in inner diameter), especially when the immersion depth was large (10 cm or 18 cm). Although an increased closing quotient of the glottis and increased closed quotient or contact quotient may reflect increased impact stress in phonation [26], it must be remembered that both EGG and high speed filming methods have their drawbacks. *CQ* from EGG has been found to get saturated while the impact stress still keeps rising [27]. The main drawback in high-speed filming is the fact that only the upper parts of the vocal folds are visible. In therapy tradition, deep bubbling (i.e. phonation through the tube immersed 10-15 cm in water) has been considered strenuous and potentially harmful for the patients, i.e. resulting in signs of vocal fatigue (tiredness of the throat and impairment of voice), if not conducted properly and for short times only, e.g. a couple of phonations in a row, see [8]. An important question is whether phonation into water, especially with a deep tube immersion can cause overloading of the vocal fold tissue due to increased contact (impact) stress during the vocal folds collisions.

This study aims to shed light on this topic by applying a physical model of the human voice production. The impact (contact) stress (*IS*) was measured directly between the synthetic vocal folds of the model in three conditions: When the model phonates on [u:] 1) without a tube, and 2) with a silicon Lax Vox tube (35 cm in length, 1 cm in inner diameter) immersed 2 cm in water, and 3) with the Lax Vox tube immersed 10 cm in water. The subglottic pressure and airflow rate ranges corresponded to normal human voice production. The measured mean subglottic and oral pressures and the peak-to-peak values of the oral pressure measured with the model are compared to those reported in [21] for 14 humans phonating into Lax Vox tube.

Methods

The model consists of vocal fold replica made of silicon and a plexiglas tube representing the vocal tract when a person articulates [u:]. Measurement set up is presented in Fig.1 and on the photographs in Fig. 2.

Figure 1.

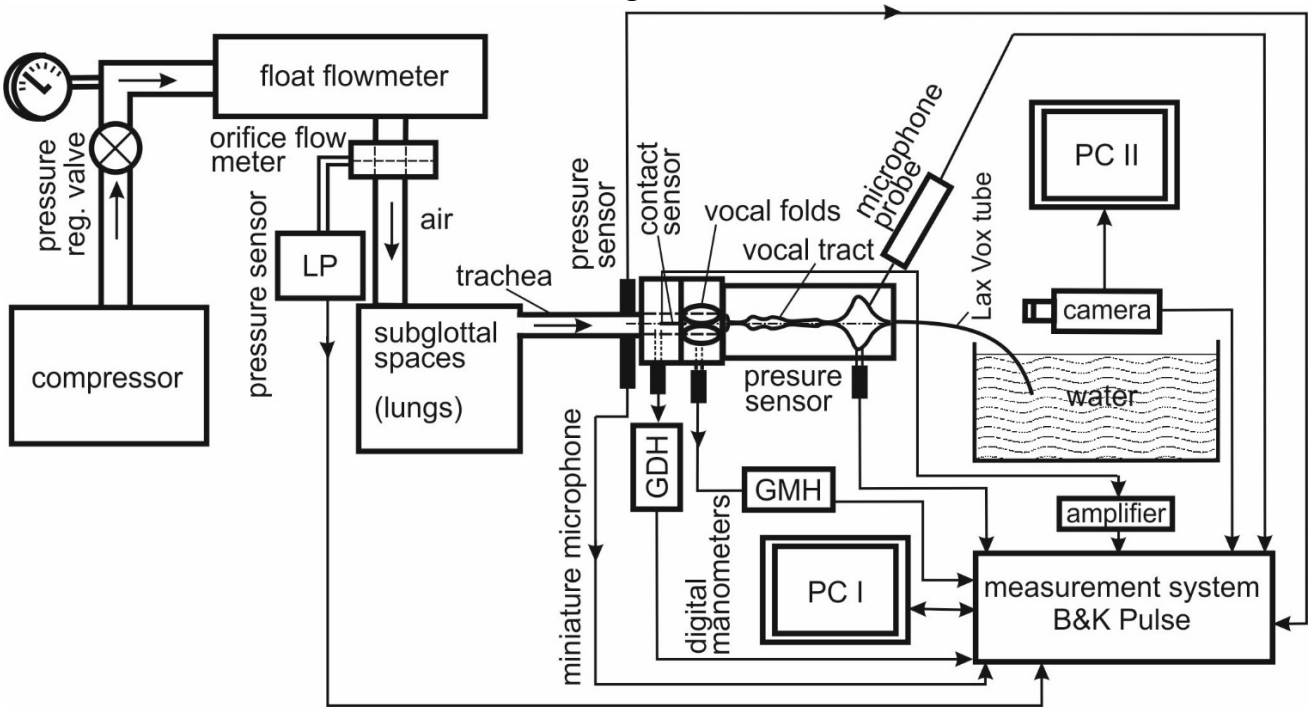
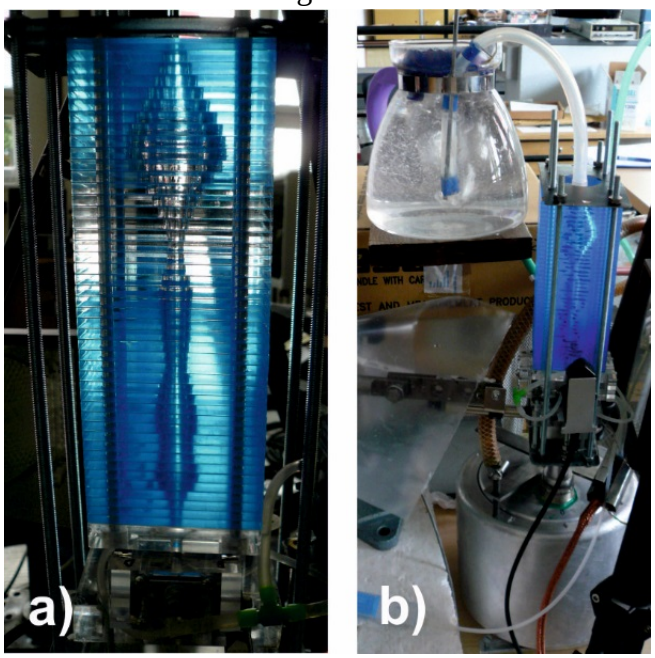


Figure 2.



The two-layered water filled silicon vocal folds replica [28,29] was excited by the airflow coming from a compressor through a regulating valve into the float flow meter, according to which the control flow rate was manually set in a given flow range. The digital orifice flow meter recorded the flow rate into the measurement system B&K Pulse. The air entered the model of subglottal spaces consisting of a simplified model of lungs and trachea where the transducer for measuring the subglottic pressure was installed. Thereafter the air flew into the part where the vocal folds were installed together with the sensor measuring the impact stress (miniature pressure transducer Precision Measurement Company model 060, range 0-350 kPa, diameter 1.5 mm, thickness 0.3 mm). The impact stress sensor was mounted on a special support positioned below the vocal folds. The

hydrostatic pressure inside the vocal folds model was regulated by a syringe in order to preset the fundamental frequency of phonation (F_0) to a fixed value. Fluctuations and the mean of the oral pressure were measured by a microphone probe B&K 4182 (range 1 Hz - 20 kHz) and by an integrated pressure semiconductor sensor NXP (Freescale MPXV5010GC6U, respectively, both installed in the oral cavity of the vocal tract model. The model for vowel [u:] was made of plexiglas, i.e. with hard walls. Vocal folds vibrations were recorded by the high speed camera NanoSense MkIII (maximum resolution 1280x1024 pixels with a camera zoom lens Nikon AF micro Nikkor 60 mm) positioned above the vocal tract model.

The high speed filming was also used to adjust the proper position of the impact stress sensor between the vibrating vocal folds as well as to make sure that the peaks in *IS* signal corresponded to vocal fold contact. It was done before the measurement of phonation on [u:] started followed directly by measurement with the Lax Vox tube.

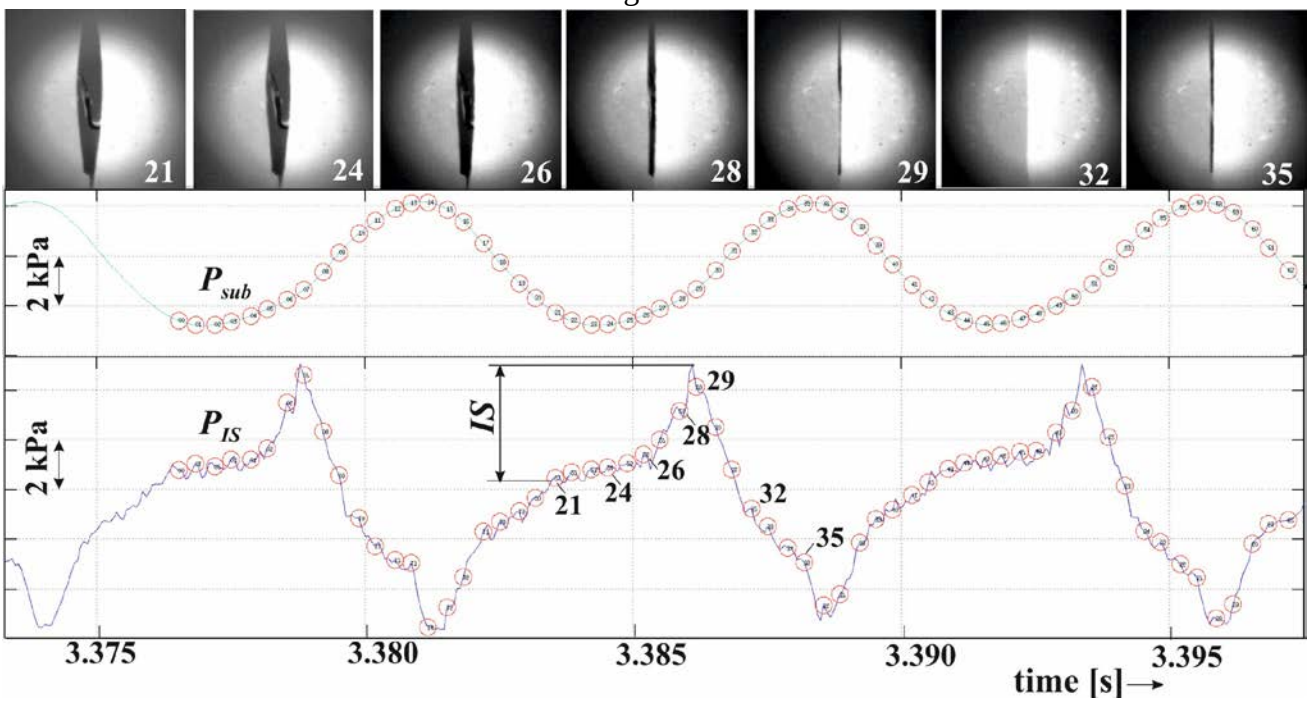
All the pressure signals were synchronously sampled and recorded by using the measurement system B&K Pulse (type 3560 C with Input/Output Controller Modules Type 7537A and 3109) controlled by a personal computer (PC I) equipped by the SW PULSE LabShop Version 10. The sampling frequency of the signals was 16.4 kHz and 3000 frames/s were synchronously recorded by the high speed camera.

The fundamental phonation frequency F_0 and the water bubbling frequency F_b , i.e. the frequency of the bubble formation at the tube end in water, were determined from the spectra of the measured oral pressure signal.

After setting the impact stress sensor in a proper position between the vocal folds for a flow rate Q slightly above the phonation threshold flow for vowel [u:] the first measurement was performed recording the pressure signals for 10 s. After saving the data in PC, the measurement was immediately repeated for higher flow rates in several steps from Q=0.15 to 0.40 l/s. Then the Lax Vox tube was joined to the vocal tract model and the measurements were immediately repeated in the same way again.

Figure 3 demonstrates how the impact stress was evaluated from the images for vocal folds vibration taken during phonation on vowel [u:]. After a maximum opening of the vocal folds, marked in the image by No 21, the magnitude of the signal P_{IS} starts to increase slowly creating a short plateau at about a minimum of the subglottic pressure P_{sub}. The signal P_{IS} is slightly increasing up to the position of the vocal folds marked by No 26 which is followed by a fast increase of P_{IS} up to the vocal folds contact, marked by No 29 in the beginning of the closed phase of the oscillation cycle. During the closed phase (images No 29–35), the subglottic pressure reaches the maximum while the signal P_{IS} is decreasing to the minimum. The distance between the maximum of P_{IS} and the level of a plateau between time instants No 21–26 was considered as the maximum of the impact stress *IS*. We note that the measurement of the impact stress during phonation into the tube with the distal end in water was more difficult, as the high speed camera was not possible to use.

Figure 3.



Results

The measured mean values of subglottic P_{sub} and oral P_{oral} pressures for all three phonation cases considered are shown in Fig. 4. All pressure values increase with the flow rate Q. As expected, the lowest values were measured for phonation on [u:] without tube and the highest values were measured for phonation through the Lax Vox tube with the distal end immersed 10 cm deep in water. The measured values were compared with the results for *in vivo* measurements published in [21]. When the flow rate in the measurement on the model was within the marked limits of about $Q \cong 0.18 - 0.27\ [\mathrm{l/s}]$, the measured P_{sub} values were within the limits (mean value $\pm$ standard deviation) found in humans for loud phonation. Similarly, in the same range of the flow rates Q, the mean values of the oral pressure measured *in vitro* for the Lax Vox tube 2 and 10 cm deep in water were comparable with the P_{oral} values measured for habitual phonation in humans.

Figure 4.

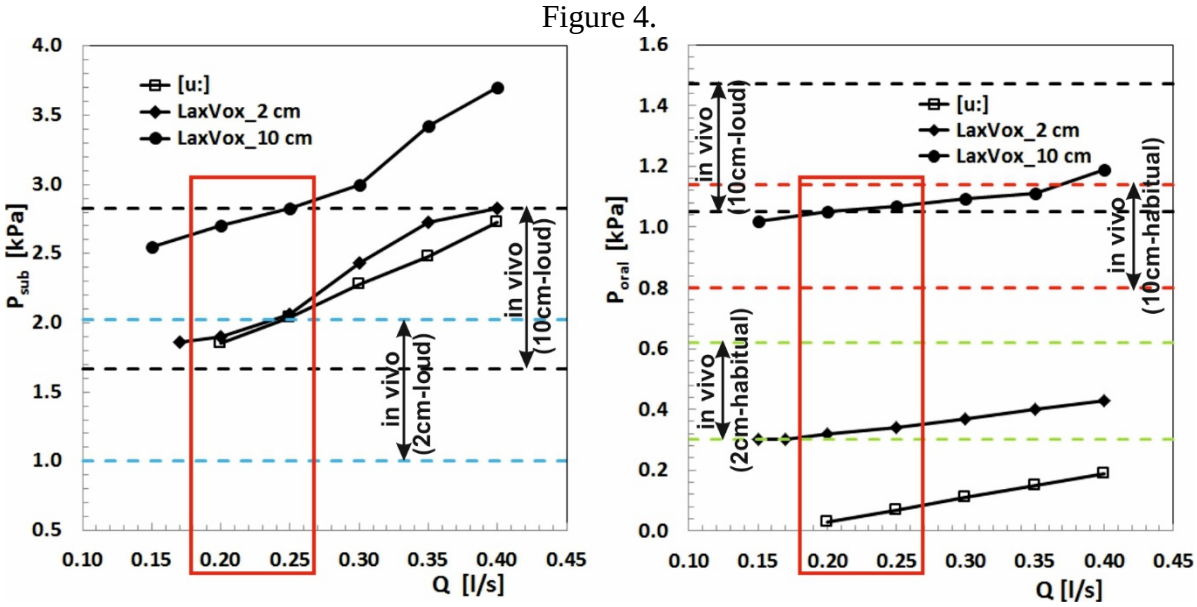


Figure 5 presents the results of the impact stress measurement on the model for phonation on vowel [u:] without tube and for phonations through the Lax Vox tube into water. The impact stress *IS* increases with the flow rate as well as with the aerodynamic power computed in trachea by multiplication of the mean subglottic pressure P_{sub} by the mean flow rate *Q*. From the results for *IS* in the marked intervals of the airflow rate and the aerodynamic power, which are comparable with the range of loud phonation for measurements in humans, ***we can conclude that the impact stress for phonation into water is smaller than for phonation on [u:].*** Because the impact stress evaluation for phonation into water was not possible to support by images of the vibrating vocal folds from the high speed camera, the results for the impact stress show a larger dispersion of the measured data for phonation into water than for phonation on [u:]. We can note that the higher *IS* measured for phonation on [u:] at the flow rate *Q*=0.25 l/s is associated with the higher transglottal pressure $P_{trans}=P_{\mathrm{sub}}-P_{\mathrm{oral}}\cong 1.96$ kPa than for phonation through the Lax Vox tube, where according to the data in Fig. 4 P_{trans}=1.72 kPa for 2cm water and P_{trans}=1.76 kPa for 10 cm water.

Figure 5.

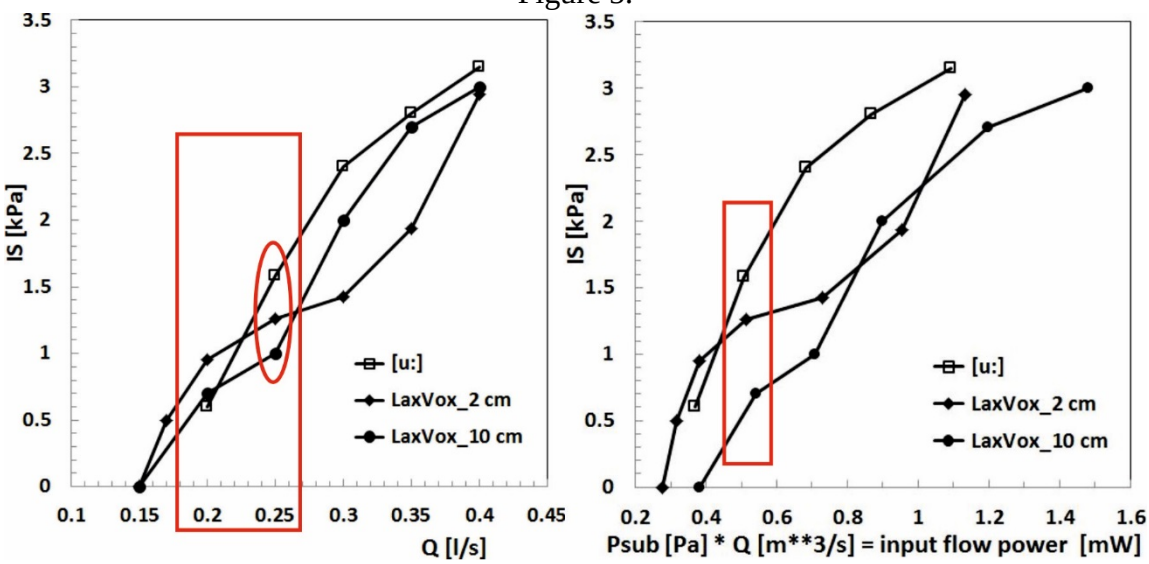
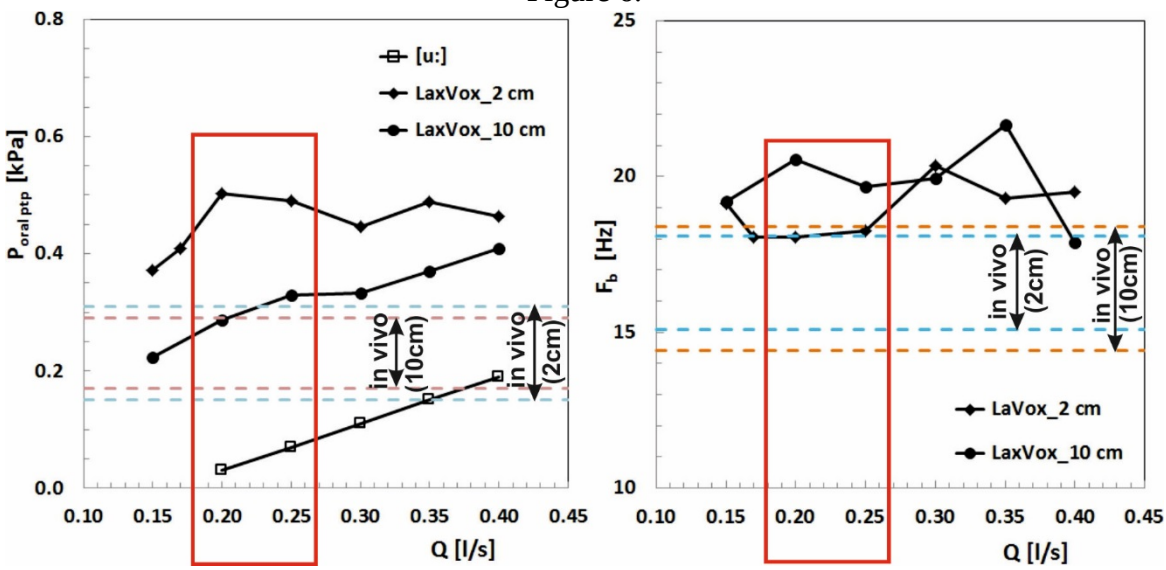


Figure 6 shows the results for the peak to peak values of the oral pressure $P_{oral\ ptp}$ and for the water bubbling frequency F_b measured on the model as well as the ranges of these quantities measured in humans. Within the marked range of the flow rates *Q*, which are comparable to the measurements in humans, the values measured *in vitro* are higher than the values measured *in vivo*. This disagreement can be explained by the yielding walls in the human vocal tract, in contrast to the hard walls in the plexiglas vocal tract model.

Figure 6.



The fundamental frequency F_0 and the flow resistance P_{oral}/Q measured in model in dependence on the flow rate Q for all considered phonations are shown in Figs. 7 and 8, respectively. The fundamental frequency decreases with the flow rate from about F_0=150 Hz to about 135 Hz. Similarly, the flow resistance in all three cases of measured phonations decreases with the flow rate. The flow resistance for phonation on the Lax Vox tube with the distal end 10 cm deep in water is markedly the highest because of the high hydrostatic pressure in water.

Figure 7.

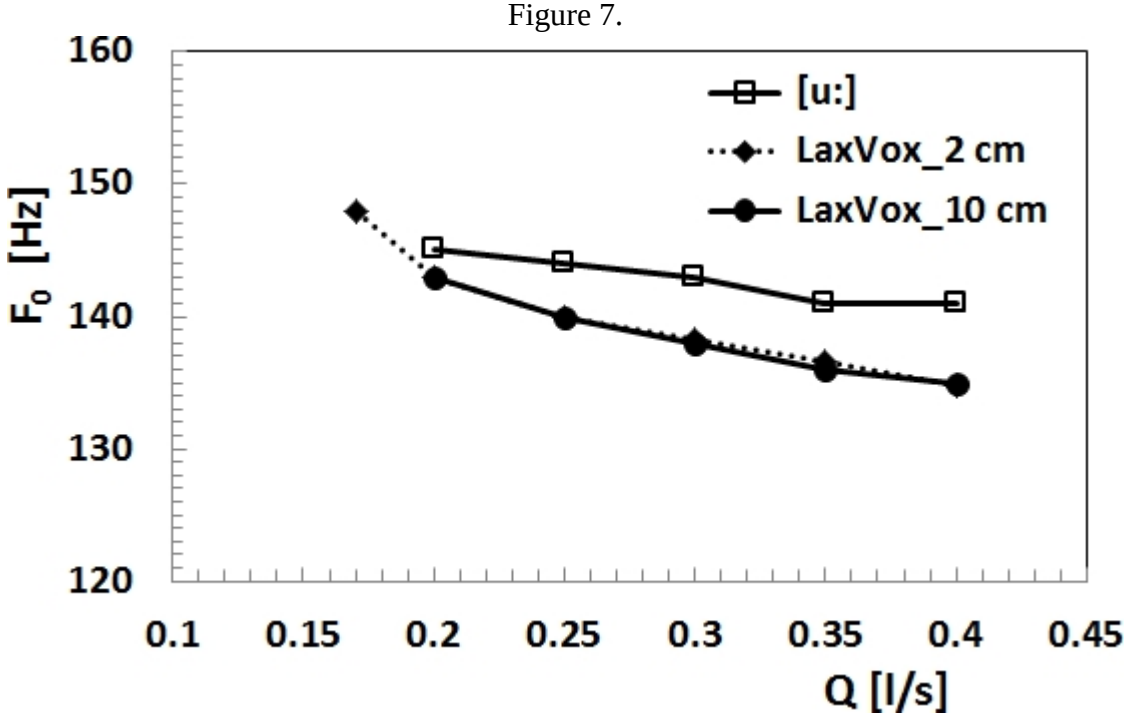
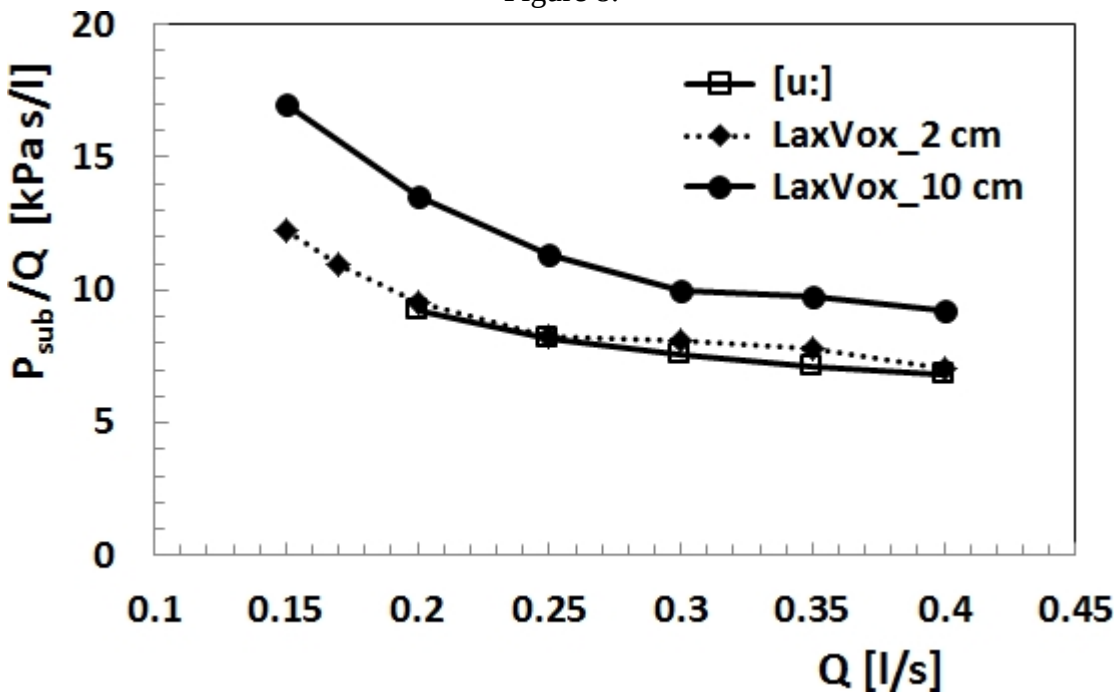


Figure 8.



Discussion

Effects of yielding walls in the human vocal tract

Some differences were found between the results of the present study and those obtained earlier in humans. The peak-to-peak amplitude values of oral pressure oscillation and the frequency of bubble formation in water (see Fig. 6) were higher in the present study compared to earlier study on humans [21]. The reason for these differences is that the vocal tract model has hard walls, while the walls of the human vocal tract are soft, yielding, see e.g. [25,30,31].

It is known that the yielding wall of the vocal tract causes a low frequency acoustic-mechanical resonance ($F_{a\text{-}m}$), which raises the acoustic resonance frequencies of the human vocal tract, i.e. the formant frequencies F_1, F_2 etc., and lowers the water bubbling frequency, see [25] and Fig. 6, respectively. Furthermore, it is well-known that the amplitudes of the pressure oscillations in the vocal tract increase if the excitation frequency is close to the resonance frequency. Thus, a question arises if such conditions for studying impact stress in water resistance therapy *in vitro* as in this paper allow comparison with the measurements in humans [21], especially as it comes to the coincidence of water bubbling frequency with the frequency of some acoustic resonance in the vocal tract.

Table 1 shows data that confirm the comparability of our results with humans. The left part of Table 1 shows the resonance frequencies $F_{a\text{-}m}$, F_1 and F_2, computed according to the recent paper of the authors [25], first by assuming the yielding walls, and then considering hard walls of the vocal tract. In the latter case, the stiffness of the vocal tract wall was considered as infinitely high. The right part of Table 1 shows the range of water bubbling frequencies F_b measured *in vivo* [21] and the ranges of F_b and fundamental frequency F_0 for *in vitro* measurements presented in Figs. 6 and 7, respectively.

It is well known that the amplitudes of the pressure oscillations increase if the excitation frequency is close to the acoustic resonance frequency. The bubbling process in humans excites the low frequency acoustical–mechanical resonance $F_{a\text{-}m}$, where according to Table 1 the difference $F_b - F_{a-m} \cong +7$ Hz was found between the bubbling frequency F_b and the theoretically estimated (computed) $F_{a\text{-}m}$, and similarly the bubbling process in the vocal tract with hard walls excites the first formant frequency F_1, where the difference $F_b - F_1 \cong -9$ Hz was found, i.e. in the first case the excitation frequency F_b

is slightly higher than the first resonance, while in the second case F_b is slightly lower than the first resonance. Therefore the conditions for measuring the impact stress in model were similar as in water resistance voice exercises applied in humans [21].

Table 1. Computed acoustical-mechanical resonance frequency $F_{a\text{-}m}$ of the vocal tract and acoustic resonance (formant) frequencies F_1, F_2 for phonation into the Lax Vox tube with distal end in water considering yielding and hard walls of the vocal tract, and the measured water bubbling frequency F_b and the fundamental phonation frequency F_0 in humans

	computed[*)]			**measured**	
vocal tract with	$F_{a\text{-}m}$ [Hz]	F_1 [Hz]	F_2 [Hz]	F_b [Hz]	F_0 [Hz]
yielding walls (in humans)	9.4	149	313	15-17[**)]	/
hard walls (in model)	/	29.2	282	18-22	135-148

*) J. Horáček, V. Radolf, A.M. Laukkanen: Low frequency mechanical resonance of the vocal tract in vocal exercises that apply tubes, *Biomedical Signal Processing and Control*, 2017.
**) J. Tyrmi, V. Radolf, J. Horáček, A.M. Laukkanen: Resonance tube or Lax Vox?, *Journal of Voice* 31 (2017) 430-437.

Table 1 also shows that the fundamental frequency F_0 was not important in case of the measurement on model, because the differences between F_0 and both formants F_1 and F_2 were higher than 100 Hz in all cases. However in humans, where the fundamental frequency F_0 is normally higher than ca 100 Hz, F_0 may be close to the first formant frequency $F_1 \cong 149$ Hz which could result in a ***double effect in the water resistance therapy*** if $F_b \cong F_{a\text{-}m}$ and coincidently $F_0 \cong F_1$.

The double effect may intensify the positive effects of water resistance therapy. On the other hand, especially if the subglottic pressure and the airflow are high (as in loud phonation) the double effect may potentially increase both impact stress and shear stress in the vocal fold tissue.

Clinical implications

The results of the present study suggest that 'water resistance therapy' implying phonation through a tube in water may decrease impact stress posed on the vocal folds, compared to ordinary vowel phonation. Therefore water resistance therapy would be less taxing than ordinary phonation. The reason for this is that increased vocal tract resistance (airpressure/airflow) during tube phonation results in an increase of mean intraglottal pressure, which reduces the contact pressure (impact stress) during vocal fold collisions [4]. This is also associated with the lower transglottal pressure as found in the present study.

It can be assumed that the altered aerodynamic situation in the vocal tract during exercising with a tube gives the trainee/patient sensations of economic voice production (adequate regulation of the expiratory airflow with respiratory muscles and adequate - neither too loose nor too tight – adduction), and that this in turn would help to learn a more economic voice production in the long run. The positive experimental findings of the effects of tube phonation support this, see e.g. [8,12,18].

Despite of the results of lower *IS*, there may be other loading effects related to water resistance therapy, possibly causing symptoms of vocal fatigue. Firstly, the activity in the adductors may increase during high airflow resistance, and this may result in tiredness of the adductor muscles. This

could cause symptoms like discomfort in the throat and possibly also deterioration of voice quality (increased breathiness) in cases where the exercising time has been excessively long and/or the water resistance has been high (e.g. in deep bubbling where the immersion depth of the tube is 10 cm or more in water). In clinical practice, deep bubbling is recommended to use for only a few short phonations at a time and mainly for patients suffering from hypofunctional dysphonia [1,2,8].

Conclusions

In this study, using a physical model of voice production, we compared the impact stress values in phonation on [u:] with the impact stress values in phonation through a Lax Vox tube in water. The comparison was performed in a corresponding range of subglottal and oral pressures as has been measured earlier in humans during water resistance exercising, see [21]. For equivalent input airflow power (aerodynamic power, i.e. subglottic pressure x airflow rate) it was shown that the impact stress can be lower for phonation on tube in water than for phonation on vowel [u:]. This suggests that water resistance exercising would be less taxing (loads the vocal fold tissue less) than ordinary phonation.

However, there can be other loading effects related to water resistance therapy. The activity in the adductor muscles may increase during high airflow resistance, which may result in tiredness of the adductors. Furthermore, when the water bubbling frequency coincides with the acoustic – mechanical resonance of the vocal tract, and especially if the fundamental frequency is simultaneously close to the first formant frequency (e.g. for female subjects) the amplitudes of oral pressure vibrations can become so high that they may result in unpleasant sensations to the subject, and potentially increase mechanical stresses in the vocal fold tissue. This warrants a further study.

Acknowledgement

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Figure captions

Figure 1. Schema of the measurement set up.

Figure 2. Photographs of the measurement set-up for (a) the vocal tract model phonating on vowel [u:], and for (b) phonation through the vocal tract prolonged by the Lax Vox tube with the distal end submerged 10 cm in water.

Figure 3. Example of impact stress evaluation from the synchronously measured subglottic pressure P_{sub} and the signal P_{IS} from the impact stress sensor using the images of the vocal folds vibration taken by the high speed camera from above the mouth orifice of the vocal tract model during phonation on [u:]. (Q=0.4 l/s, P_{sub}=2465 Pa, F_0=137 Hz)

Figure 4. Mean values of subglottic P_{sub} (left) and oral P_{oral} (right) pressures, measured on the model for phonation on [u:] and with the Lax Vox tube with the distal end immersed 2 cm and 10 cm deep in water, compared with similar measurement in humans, see Tyrmi et al. 2017 [21]. (On *x*-axis: airflow rate Q measured in liters per second [l/s]; on *y*-axes: mean subglottic pressure, P_{sub}, and mean oral pressure, $P_{oral,}$ measured in kilopascals [kPa].)

Figure 5. Impact stress *IS* measured on the model for phonation on [u:] and for phonation with the Lax Vox tube with the distal end immersed 2 cm and 10 cm deep in water as a function of the flow rate Q (left) and the aerodynamic power (right). (On *x*-axes: airflow rate Q measured in liters per second [l/s] and aerodynamic power measured in miliwatts [mW], respectively; on *y*-axes: impact stress *IS* measured in kilopascals [kPa].)

Figure 6. Peak-to-peak values of the oral pressure $P_{oral\ ptp}$ (left) and the water bubbling frequency F_b (right) for phonation on [u:] without the Lax Vox tube and for phonation through the Lax Vox tube with the distal end immersed 2 cm and 10 cm in water. Data obtained on model are compared with the measurements in humans, see Tyrmi et al. 2017 [21]. (On *x*-axes: airflow rate Q measured in liters per second [l/s]; on *y*-axes: peak-to-peak oral pressure $P_{oral\ ptp}$ in kilopascals (?) [kPa] and bubbling frequency F_b measured in Hertz [Hz], respectively.)

Figure 7. Fundamental phonation frequency F_0 measured in model for phonation on the vowel [u:] and the Lax Vox tube with the distal end 2 and 10 cm deep in water. (On *x*-axis: airflow rate Q measured in liters per second [l/s]; on *y*-axis: fundamental frequency F_0 measured in Hertz [Hz].)

Figure 8. Flow resistance measured in model for phonation on the vowel [u:] and the Lax Vox tube with the distal end 2 and 10 cm deep in water. (On *x*-axis: airflow rate Q measured in liters per second [l/s]; on *y*-axis: mean subglottic pressure P_{sub} measured in kilopascals [kPa] divided by airflow rate Q measured in liters per second [l/s].)