Impact Stress in Water Resistance Voice Therapy: A Physical Modeling Study

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Summary: 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: This is an experimental modeling study.
Methods: A model with three-layer silicone vocal fold replica and a plexiglass, MK Plexi, Prague 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 immersed 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 with [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 (eg, 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 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 50 years ago.[1,2] Afterward, many studies have been conducted describing the method itself[3−8] and 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," 35 cm in length, 1−1.2 cm in inner diameter.[5] Resonance tube is recommended to immerse 2 cm in water for the treatment of, for example, hyperfunctional voice disorders, whereas a deeper immersion, up to 10 cm or even 15 cm in water, has been used to treat hypofunction, for example, unilateral vocal fold paresis.[8] Lax Vox has been recommended to be submerged 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 for example References 3,10). It is well known that the depth of immersion of the tube in water regulates the airflow resistance (see for example References 15,21,23,24). Increased airflow resistance increases intraglottal air pressure and thus tends to reduce collision between the vocal folds during phonation.[3,4,6] Modeling results and some electroglottographic (EGG) observations on humans support this (see for example References 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 and Laukkanen[20] also showed that in some cases the contact quotient was higher in subjects phonating through a tube immersed 10 cm in water compared with 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 or 18 cm). Although an increased closing quotient of the glottis and an increased closed quotient or contact quotient may reflect increased impact stress (*IS*) in phonation,[25] 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 *IS* still keeps rising.[26] 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 (ie, phonation through the tube immersed 10−15 cm in water) has been considered strenuous and potentially harmful for the patients, that is,

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resulting in signs of vocal fatigue (tiredness of the throat and impairment of voice), if not conducted properly and for short times only, for example, a couple of phonations in a row (see Simberg and Laine[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 fold 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, (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 with those reported in Tyrmi et al[21] for 14 humans phonating into Lax Vox tube.

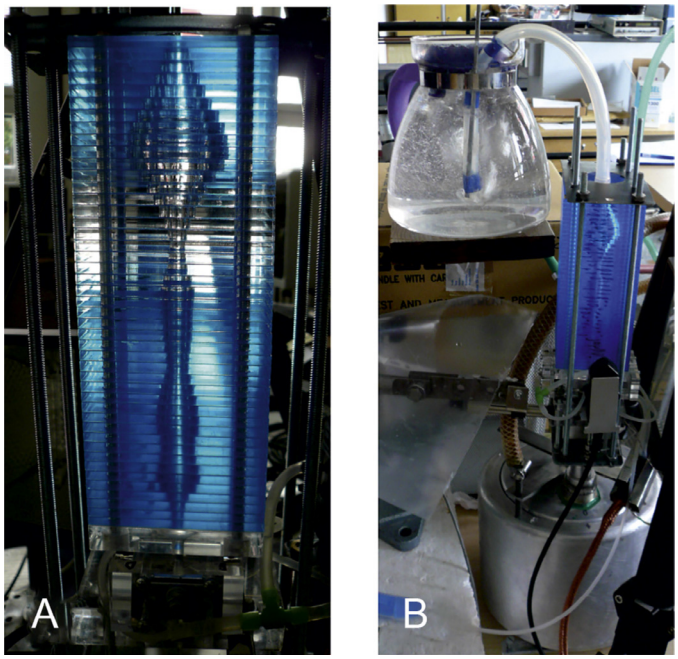

FIGURE 2. Photographs of the measurement setup for (**A**) the vocal tract model phonating on vowel [u:] and for (**B**) phonation through the vocal tract prolonged by the Lax Vox tube with the distal end submerged 10 cm in water.

METHODS

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

The two-layer water-filled silicone vocal fold replica[27,28] 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 (Brüel & Kjær Sound & Vibration Measurement, Denmark). 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 *IS* (miniature pressure transducer Precision Measurement Company (Ann Arbor, USA) model 060, range 0−350 kPa, diameter 1.5 mm, thickness 0.3 mm). The *IS* sensor was mounted on a special support positioned below the vocal folds. The hydrostatic pressure inside the vocal fold model was regulated by a syringe 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 10 kPa NXP (FREESCALE)

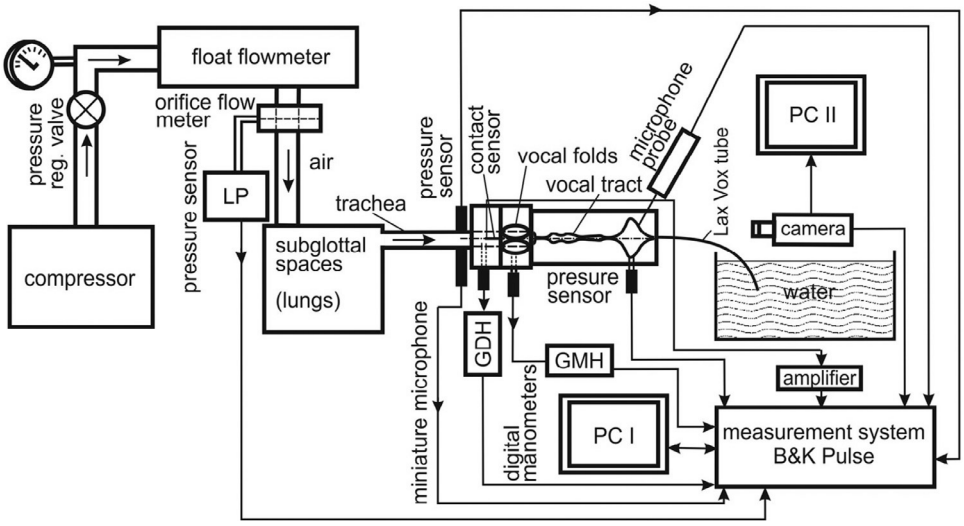

FIGURE 1. Schema of the measurement setup.

type MPXV5010GC6U (Eindhoven, The Netherlands), respectively, both installed in the oral cavity of the vocal tract model. The model for vowel [u:] was made of plexiglass, that is, with hard walls. Vocal fold vibrations were recorded by the high-speed camera NanoSense MkIII (maximum resolution 1280×1024 pixels with a camera zoom lens Nikon Inc. [Melville, NY, USA] 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 *IS* sensor between the vibrating vocal folds, as well as to ensure 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 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 (Brüel & Kjær Sound & Vibration Measurement, Denmark). The sampling frequency of the signals was 16.4 kHz, and 3000 frames/s was synchronously recorded by the high-speed camera.

The fundamental phonation frequency (F_0) and the water bubbling frequency (F_b), that is, 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 *IS* 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 seconds. 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.

Figure 3 demonstrates how the *IS* was evaluated from the images for vocal fold 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 *IS*. We note that the measurement of the *IS* during phonation into the tube with the distal end in water was more difficult, as the high-speed camera was not possible to be used.

RESULTS

The measured mean values of subglottic P_{sub} and oral P_{oral} pressures for all three phonation cases considered are shown in Figure 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.[21] When the flow rate in the measurement on the model was within the marked limits of about $Q \cong 0.18-0.27$[L/s], the measured P_{sub} values were within

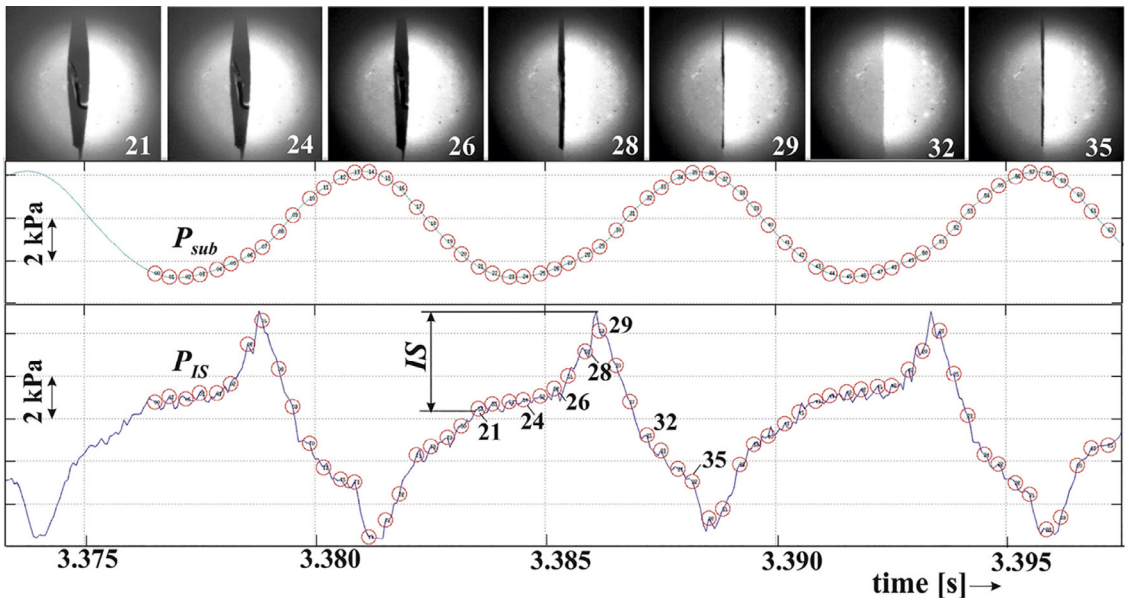

FIGURE 3. Example of impact stress evaluation from the synchronously measured subglottic pressure P_{sub} and the signal P_{IS} from the impact stress sensor using the images of the vocal fold vibration taken by the high-speed camera from above the mouth orifice of the vocal tract model during phonation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

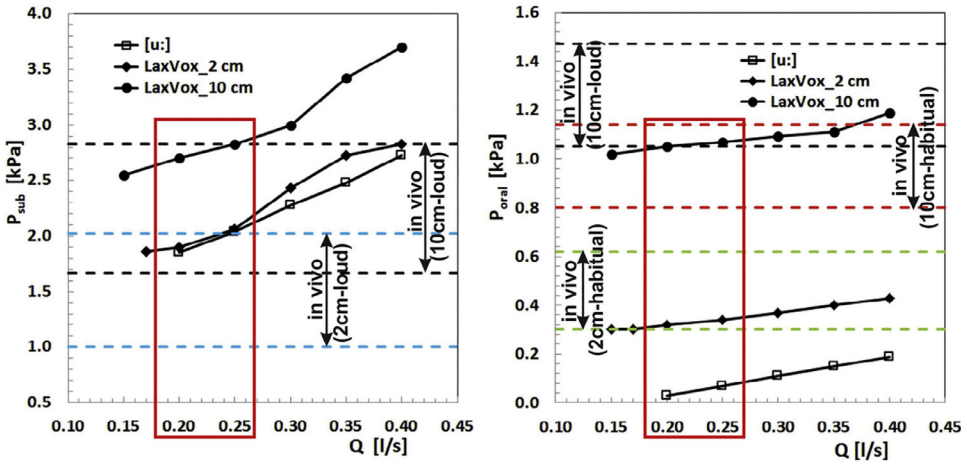

FIGURE 4. Mean values of subglottic P_{sub} (left) and oral P_{oral} (right) pressures, measured on the model for phonation on [u:] and with the Lax Vox tube with the distal end immersed 2 and 10 cm deep in water, compared with similar measurement in humans (see Tyrmi et al[21]). (On the *x*-axis: airflow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

the limits (mean value ± 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 5 presents the results of the *IS* measurement on the model for phonation on vowel [u:] without tube and for phonations through the Lax Vox tube into water. The *IS* increases with the flow rate and 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 *IS* evaluation for phonation into water was not possible to be supported by images of the vibrating vocal folds from the high-speed camera, the results for the *IS* 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 $P_{trans} = P_{sub} - P_{oral} \cong 1.96$ kPa than for phonation through the Lax Vox tube, where according to the data in Figure 4, $P_{trans} = 1.72$ kPa for 2 cm water and $P_{trans} = 1.76$ kPa for 10 cm water.

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 with the measurements in humans, the values measured *in vitro* are higher than the values measured *in vivo*. This disagreement

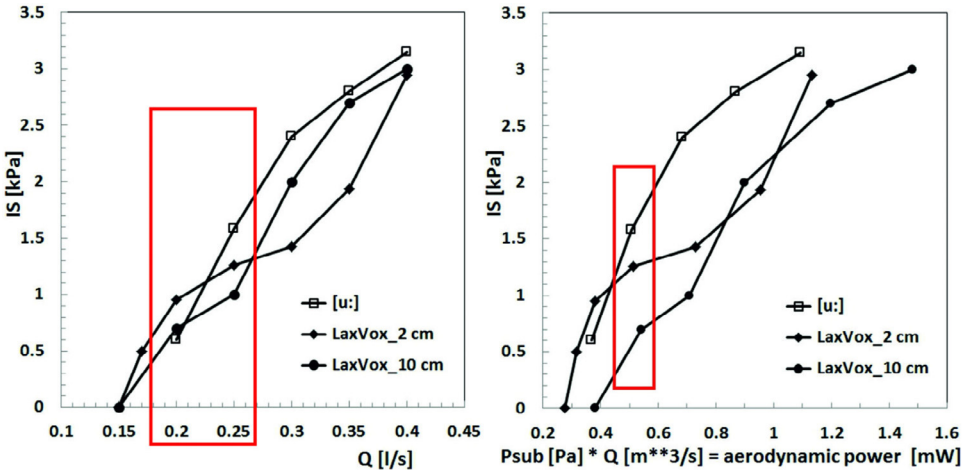

FIGURE 5. Impact stress (*IS*) measured on the model for phonation on [u:] and for phonation with the Lax Vox tube with the distal end immersed 2 and 10 cm deep in water as a function of the flow rate Q (left) and the aerodynamic power (right). (On the *x*-axis: airflow rate Q measured in liters per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

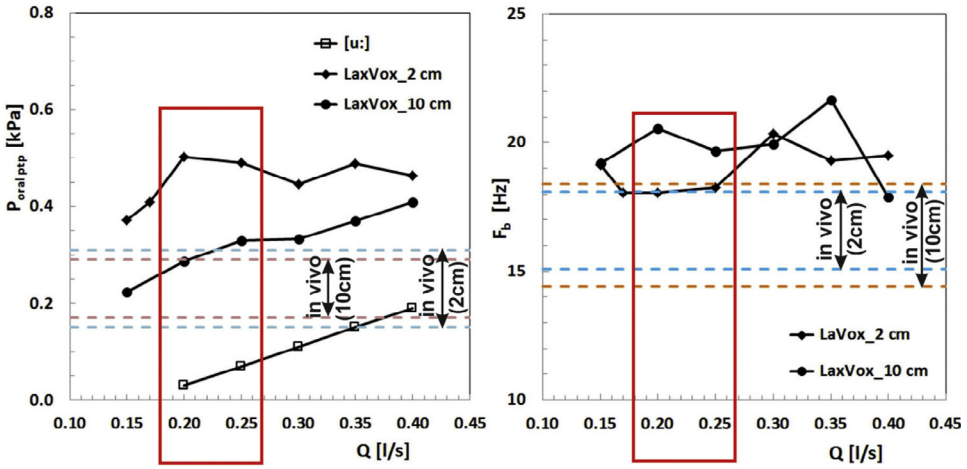

FIGURE 6. Peak-to-peak values of the oral pressure $P_{oral\ ptp}$ (left) and the water bubbling frequency F_b (right) for phonation on [u:] without the Lax Vox tube and for phonation through the Lax Vox tube with the distal end immersed 2 and 10 cm in water. Data obtained on model are compared with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

can be explained by the yielding walls in the human vocal tract, in contrast to the hard walls in the plexiglass vocal tract model.

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 Figures 7 and 8, respectively. The fundamental frequency decreases with the flow rate from about $F_0 = 150$ 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.

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 Figure 6) were higher in the present study compared with earlier study on humans.[21] The reason for these differences is that the vocal tract model has hard walls, whereas the walls of the human vocal tract are soft, yielding, see for example References 29–31.

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, that is, the formant frequencies F_1, F_2, etc, and lowers the water bubbling frequency, see Horáček et al[29] and Figure 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 *IS* in water resistance therapy *in vitro* as in this paper allow comparison with the measurements in humans,[21] especially as

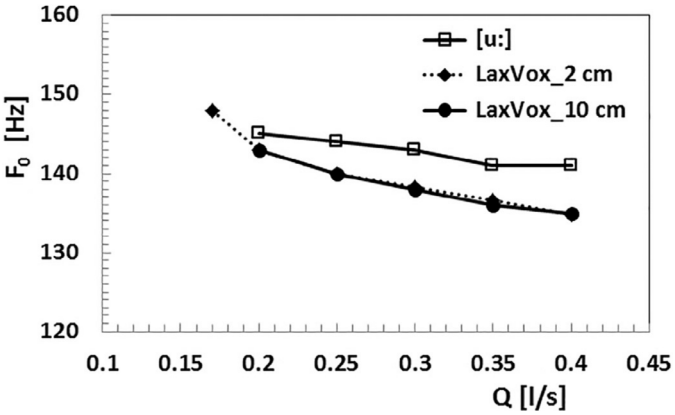

FIGURE 7. Fundamental phonation frequency (F_0) measured in model for phonation on the vowel [u:] and the Lax Vox tube with the distal end 2 and 10 cm deep in water. (On the *x*-axis: airflow rate Q measured in liters per second [L/s]; on the *y*-axis: fundamental frequency F_0 measured in Hertz [Hz].)

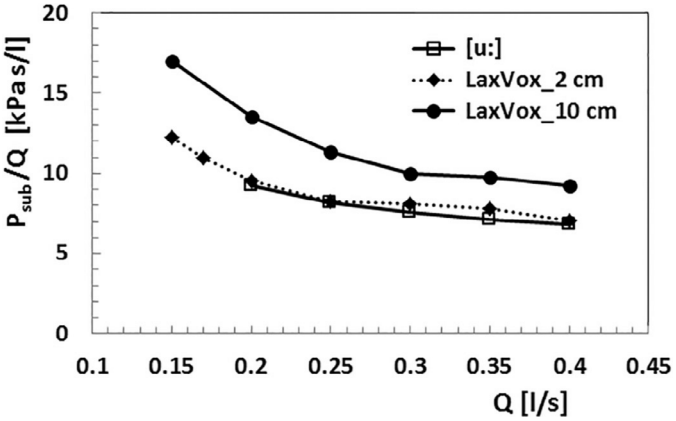

FIGURE 8. Flow resistance measured in model for phonation on the vowel [u:] and the Lax Vox tube with the distal end 2 and 10 cm deep in water. (On the *x*-axis: airflow rate Q measured in liters per second [L/s]; on the *y*-axis: mean subglottic pressure P_{sub} measured in kilopascals [kPa] divided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

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)

	Computed*			Measured (a) in Humans, (b) on Physical Model	
Vocal Tract With	$F_{a\text{-}m}$ (Hz)	F_1 (Hz)	F_2 (Hz)	F_b (Hz)	F_0 (Hz)
Yielding walls assumed (as in humans)	9.4	149	313	(a) 15–17†	/
Hard walls assumed (as in physical model)	/	29.2	282	(b) 18–22	135–148

*Horáček et al.[29]
†Tyrmi et al.[21]

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,[29] 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 Figures 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 $F_b - F_{a\text{-}m} \cong +7$Hz wasfound 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, that is, in the first case the excitation frequency F_b is slightly higher than the first resonance, whereas in the second case, F_b is slightly lower than the first resonance. Therefore, the conditions for measuring the *IS* in model were similar as in water resistance voice exercises applied in humans.[21]

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. In contrast, especially if the subglottic pressure and the airflow are high (as in loud phonation), the double effect may potentially increase both *IS* 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 *IS* posed on the vocal folds, compared with ordinary vowel phonation. Therefore, water resistance therapy would be less taxing than ordinary phonation. The reason for this is that increased vocal tract resistance (air pressure or airflow) during tube phonation results in an increase of mean intraglottal pressure, which reduces the contact pressure (*IS*) 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 or 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 for example References 8,12,18).

Despite the results of lower *IS*, there may be other loading effects related to water resistance therapy, possibly causing symptoms of vocal fatigue. First, 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 such as discomfort in the throat and possibly also deterioration of voice quality (increased breathiness) in cases where the exercising time has been excessively long and the water resistance has been high (eg, 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 *IS* values in phonation on [u:] with the *IS* 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 exercise (see Tyrmi and Laukkanen[21]). For equivalent input airflow power (aerodynamic power, ie, subglottic pressure × airflow rate), it was shown that the *IS* 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 (eg, 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.

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