MODELLING OF THE LARYNGECTOMEE SUBSTITUTE VOICE

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Abstract: A bio-mechanical model is derived which describes the fundamental principles of laryngectomie substitute voice production. Within the model the substitute voice generator (PE segment) is modelled as an elastic tube which is set into vibrations by streaming air. The model bases on the well known two-mass-model by Ishizaka and Flanagan (1972) which has been successfully used to describe regular phonation. The morphology of the PE segment is considered by several two-mass-models which are orbitally coupled with spring and damping elements. The main parameters which affect oscillation are vibrating masses, muscle tensions and lung pressure. Within the model, the time dependent minimum aperture serves as measure of PE segment deformations. The performance of the PE-Model is demonstrated by adapting the PE-Model to experimental PE segment vibrations which are extracted from high-speed sequences.

Keywords: Substitute Voice, High-Speed-Recording, Two-Mass-Model, PE-Model.

I. INTRODUCTION

Therapy of cancer of the throat may require a surgical excision of the larynx which results into the loss of voice. During a so-called total laryngectomy, trachea and esophagus are separated in order to prevent uncontrolled mixing of breathing and swallowing [1]. Breathing is maintained by suturing the trachea into the frontal skin of the neck (tracheostoma). In order to achieve voice rehabilitation a substitute voice generating element has to undertake the task of the excised larynx. State of the art therapy is the insertion of a silicon shunt valve which reconnects the separated trachea and esophagus and establishes an unidirectional connection from the trachea to the esophagus [2]. When closing the tracheostoma during expiration, air passes through the voice prosthesis into the esophagus. The airflow excites vibrations of soft tissue at the upper esophagus sphincter, i.e. pharyngeal-esophageal segment (PE segment). These tissue vibrations modulate the airstream which poses as substitute voice signal (tracheoesophageal voice production). The anatomy of the PE segment consists of a mucosal coated ring-shaped muscle structure. The aim of this work is to introduce a bio-mechanical model of the PE segment (PE-Model) which describes PE segment dynamics in order to gain insight into the voice generating process.

II. METHODOLOGY

A. Analyzing PE Vibrations in High-Speed Sequences

High-speed recordings are performed during phonation using an endoscope coupled with a digital high-speed camera which allows the observation of PE segment vibrations in real-time. The patients are instructed to articulate the vowel /a/ in a ‘comfortable’ way. The frame rate of the high-speed system is 3704 Hz while the resolution of the CCD-array is 128 x 64 pixel. Simultaneously, the acoustic signal is recorded. For two high-speed recordings the tissue vibrations are quantitatively analyzed during a time interval of 95 ms using an image processing algorithm [3, 4]. The size and shape of the pseudoglottis a(t), which is determined by the algorithm, serves as measure for PE segment deformations.

B. Model of the Pharyngeal Esophageal Segment

The principle properties of voice production of laryngeal (vocal folds) and tracheoesophageal phonation (PE segment) are similar to each other. In both cases tissue vibrations are excited by aerodynamic forces which are caused by airflow. The aerodynamic forces can be described by the Bernoulli law while the myoelastic tissue vibrations follow bio-mechanics. Therefore, the here proposed model of substitute voice generation is derived from the model of vocal folds by Steinkecke and Herzl [5] which bases on the Two-Mass-Model (2MM) developed by Ishizaka and Flanagan [6]. Though the 2MM contains a lot of simplifications concerning both the myoelastic and the aerodynamic part, it allows the description of the most important features of vocal fold dynamics. It has successfully been used to study vocal fold vibrations in voice production.
As first approximation the upper part of the esophagus is regarded as a flexible tube. The morphology is integrated into the PE-Model by placing several 2MM orbitaly onto a horizontal circle. The center of the circle is regarded as point of origin of a cartesian coordinate system. Each 2MM is orientated to this point of origin. As the esophagus is a closed elastic tube the 2MMs are horizontally connected to each other. Therefore, adjacent 2MMs are coupled by additional spring and damping elements $k^h, r^h$. Fig. 1 shows the PE-Model with circular geometry, comprising eight masses per plane. Since the horizontal coupling extends the degree of freedom each mass is capable to move within the entire (x,y)-plane. The PE-Model is described by the following differential equation:

$$0 = m_{s,i} \ddot{x}_{s,i} + r_{s,i} \dot{x}_{s,i} + k_{s,i} |\Delta x_{s,i}| u_{s,i}^0 + k_{s,i}^h |\Delta x_{s,i}^h| u_{s,i}^0 + k_{s,i}^r |\Delta x_{s,i}^r| u_{s,i}^0 + r_{s,i} (x_{s,i} - x_{s+1,i}) + F^D_{s,i} + F^F_{s,i} + F^H_{s,i}. \quad (1)$$

The indices $i$ denote the number of masses $m_{s,i}$ within the lower ($s = 1$) and upper ($s = 2$) plane. The differential equation contains tissue properties of the PE segment, i.e. masses $m_{s,i}$, stiffness $k_{s,i}$, $k_{s,i}^h$, $k_{s,i}^r$, and damping coefficients $r_{s,i}$, $r_{s,i}^h$, $r_{s,i}^r$. $x_{s,i}$ denote the position of the masses $m_{s,i}$ within the cartesian coordinate system. Spring length variations in respect to the rest position of the masses $x_{s,i}^0$ are expressed by the supplement $\Delta$. The unit vectors $u_{s,i}$ indicate the directions of spring and damping elements of mass $m_{s,i}$ and are illustrated in Fig. 2.

The driving forces $F^D_{s,i}$ result from pressure variations within the PE segment and depend on height of the lower plane $\delta_1$ the area ratio of the minimal area $a_{min}$ of both planes and the area of the lower plane $a_1$:

$$F^D_{s,i} = P_L \cdot L_{1,i} \cdot \delta_1 \cdot (1 - \frac{a_{min}}{a_1})^2 \cdot u_{s,i}^D. \quad (2)$$

The directions $u_{s,i}^D$ of the driving forces $F^D_{s,i}$ are defined in Fig. 3. The influence of colliding tissue is considered by additional spring constants $k_{s,i}^c$. Fig. 4 illustrates the collision force $F^c_{s,i,n}$ for a single impact. Collisions occur when a mass $m_{s,i}$ collides with a coupling spring of two adjacent masses $m_{s,j}, m_{s,j+1}$.
are considered by additional coupling elements $k_{s,i,n}^h$ and $r_{s,i,n}^h$ ($n = 1, 2$ denotes the left and right coupling string and damping element of mass $m_{s,i}$).

III. RESULTS

A. Adjustment to high-speed recordings

The performance of the PE-Model is exemplarily demonstrated by modifying the parameters $P_L$ and $k_{s,i}$ to match the PE-Model to observable pseudoglottis deformations $a(t)$. These pseudoglottis deformations are extracted from the high-speed sequences HS-I and HS-II which had been recorded during the examination of two different laryngectomies.

![Figure 3: Graphical definition of the driving Force $F_{s,i}^D$ and its corresponding unit vector $u_{s,i}^0$. The damping elements are not illustrated.](image)

At impact a collision spring $k_{s,i,j}^c$ acts along the dotted line in direction of $u_{s,i}^0$ which results into the impact force

$$F_{s,i}^c = k_{s,i,j}^c \sum_{j=1}^{n-2} \gamma_{s,i,j} \cdot u_{s,i}^0.$$  \hspace{1cm} (3)

![Figure 4: Graphical definition of a collision between the mass $m_{s,i}$ and the spring of the adjacent masses $m_{s,j}$ and $m_{s,j+1}$. The penetration depth $\gamma_{s,i,j}$ and the spring $k_{s,i,j}^c$ indicate the strength of the collision. The damping elements are not illustrated.](image)

Finally, the horizontal coupling forces

$$F_{s,i}^H = \sum_{n=1}^{2} r_{s,i,n}^h (x_{s,i} - x_{s,n}) + k_{s,i,n}^h |\Delta x_{s,i,n}^h| u_{s,i,n}^h.$$  \hspace{1cm} (4)

![Figure 5: Top: The solid lines show the experimental PE deformations $a(t)$ while the dotted lines show the simulation results $a_{min}(t)$. Middle: Amplitude spectra of experimental PE deformations $A(f)$. Bottom: Amplitude spectra of the modelled PE deformations $A_{min}(f)$.](image)

The parameters of each 2MM within the PE-Model are initially derived by dividing the standard parameter set of Ishizaka and Flanagan [6] by the number of single 2MM used within the PE-Model.
Thus, the dynamic properties of the PE-Modell do not depend on the number of 2MM within the PE-Model. In both cases the horizontal coupling is defined as

\[
\begin{align*}
    k_{s,i,n}^h & := (k_{s,i} + k_{s,i\pm 1}) \cdot 0.0854 \\
    r_{s,i,n}^h & := (r_{s,i} + r_{s,i\pm 1}) \cdot 0.025
\end{align*}
\]  

(5)

while the springs and masses within different planes show the following relation

\[
\begin{align*}
    k_{s+1,i} & := 0.1 \cdot k_{s,i} \\
    m_{s+1,i} & := 0.2 \cdot m_{s,i}
\end{align*}
\]  

Finally, the lung pressure \( P_L \) and the spring constants \( k_{s,i} \) are manually modified. For HS-I the determined lung pressure is \( P_L = 42.5 \text{ cm H}_2\text{O} \) while the spring constants are \( k_{1,i} = 0.0153 \). For HS-II the lung pressure is \( P_L = 31.1 \text{ cm H}_2\text{O} \) while the spring constants are \( k_{1,i} = 0.0077 \). The adaptation results obtained with the two modified parameter sets are shown in Fig. 5. Within the upper two graphs the dotted lines show the simulated PE deformations represented by \( \epsilon_{s,i}(t) \) while the solid lines show the experimental PE deformations \( \epsilon(t) \) extracted from the high-speed recordings during a time interval of 95 ms. The differences between the curves are hardly visible, since the simulated PE vibrations match very precisely the experimentally extracted PE deformations. The amplitude spectra \( A(f) \) of both experimental and simulated PE dynamics are shown. The constant components in Fourier-Space are eliminated. Within each spectra characteristic frequencies \( f_i \) can clearly be identified. Besides the fundamental frequencies of 190 Hz and 127 Hz the PE-Model simulates successfully the characteristic frequencies \( f_i \).

IV. DISCUSSION

This paper describes a bio-mechanical model which allows the simulation of the substitute voice generating process. Within the PE-Model the two dimensional morphology of the PE segment is considered by coupling orbitally multiple harmonic oscillators with additional spring and damping elements. The PE-Model can successfully be used to model the fundamental characteristics of PE segment vibrations. This is demonstrated by adapting the PE-Model manually to experimental PE segment vibrations which had been extracted from different high-speed recordings. The simulation results showed identical vibratory characteristics as experimental PE segment vibrations. In summary, this work is the first approach to model PE segment dynamics in order to gain insight into the substitute voice generating process. In a further project a fully automatic adaptation and optimization of the PE-Model to match experimental PE segment vibrations is intended [7] in order to derive physiological parameters of the PE segment. Furthermore, the PE-Model shall be used to investigate the correlation between PE-dynamics and substitute voice quality.

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REFERENCES


