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PHYSICAL MODELS OF THE LARYNX SOURCE

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The development of objective acoustic methods in the medical diagnostics of several speech disorders and the resulting clinical applications in laryngology and phoniatry call for a versatile physical model of the human larynx source. Such a model should simulate the physiological structure and the characteristics of the natural, i.e. biological laryngeal system from both the phenomenological and the quantitative points of view. In the present paper the performance and the general characteristics of the human larynx source are briefly described and its acoustic parameters are defined. Special attention is paid to the physical meaning of these parameters and to the parallels which exist between them and the anatomical structure of the biological system. On the basis of a few rationally motivated simplifying assumptions the mechanical model of the human larynx source and its equivalent electrical analogue circuit are described and discussed. Special attention is paid to the two-mass model which is very convenient for diagnostic purposes. The physical interpretation of the model's acoustic parameters is given and its mathematical description is formulated, the latter being expressed in the form of two sets of differential equations, describing air-flow and mass-movement, respectively. Finally, the convenience and the usefulness of the model in application to laryngological and phoniatric diagnostics of larynx disorders is briefly discussed and validated.

1. Introduction

The biological larynx generator, which is the only source of energy in the articulation of vowels and nasal consonants, and the main source of energy in the articulation of all remaining voiced speech sounds, plays a fundamental rôle in the process of conveying information by means of speech. It affects both the *linguistic information* which is responsible for the phonemic and the semantic content of the conveyed message, and the *personal information* which describes the individual features of the speaker's voice. Since the larynx source not only enables the correct articulation of all voiced sounds, but also is responsible for several suprasegmental prosodic features of speech, such as intonation, melody, accent and duration, it is evident that a knowledge of the mechanism of its action as well as of the relations that exist between its acoustic parameters and the anatomical structure of the source itself is necessary both for programmed speech synthesis in technical systems, e.g. in computers, and in medical diagnostics and/or in rehabilitation of speech disorders. The latter application is of particular importance since the pathology and the anomalies of the larynx constitute a very high proportion of problems in laryngology and phoniatry. It concerns both congenital and developmental defects, defects caused by professional diseases and those arising from necessary surgical intervention, viz. laryngotomy and tracheotomy.

Acoustic diagnostic methods in laryngology and phoniatry, which are based, generally speaking, on an analysis of the information content of the speech signal as the final and natural output from the speech—organs, have recently become commonly used in clinical practice (cf.e.g. [14]). They not only assist, but in some cases are even superior to the prior classical methods, e.g. laryngoscopy, stroboscopy, electromyography and radiography etc., since:

– they are used under normal conditions of phonation and articulation,

– they neither need any surgical tool or foreign substances to be introduced into the speech organ, nor have to be aided by any dangerous and painful, intrusive procedures,

– they enable the real-time visualization of the acoustic parameters of the speech signal, e.g. on TV monitor. They are thus especially convenient during rehabilitation since the hearing disorders, usually associated with speech disorders, may be compensated by the additional and auxiliary information channel of the sight organ.

Acoustical methods consist, generally, in the measurement and analysis of those acoustical parameters of the speech signal which describe the internal structure of the respective sound source, in laryngeal diagnostics the larynx generator [13]. The development and application of these methods in clinical practice must be supported by a versatile physical model of the human larynx source, a model that can simulate the action and the characteristics of the natural biological system not only in qualitative, but also in quantitative terms.

2. The mechanism of action and the general characteristics of the larynx source

2.1. *The principles of action and the physical parameters of the larynx source.* Fig. 1 represents the anatomical structure of the human larynx. The larynx source may be considered from a physical point of view as an aerodynamic oscillator whose behaviour is determined by some physical parameters (such as subglottal pressure and mechano-acoustical or structural constants of the vocal folds) and also to some extent depends on the acoustic load represented by the input impedance of the vocal tract. In the general case this tract consists of the pharynx, mouth and nasal cavities. The action of this oscillator must be considered in close connection with that of the subglottal part of the human respiratory system, which is shown in Fig. 2 in the form of a simplified

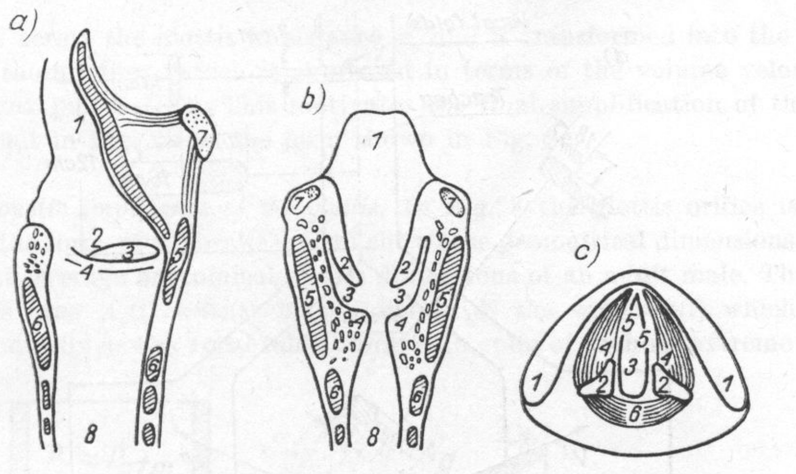

Fig. 1. Anatomical structure of the human larynx [16]

a) profile cross-section: 1 – epiglottis, 2 – false vocal fold, 3 – ventricle of Morgagni, 4 – vocal fold, 5 – thyroid cartilage, 6 – cricoid cartilage, 7 – hyoid bone, 8 – trachea; b) frontal cross-section (notations as above); c) horizontal cross-section: 1 – thyroid cartilage, 2 – arytenoid cartilages, 3 – glottis, 4 – vocalis muscles, 5 – vocal ligaments, 6 – transverse arytenoid muscles

physical model and its two equivalent electrical circuits in the impedance-type system of analogy.

The vocal folds, having a definite mass m, stiffness s and loss-resistance r, are set in forced vibrations by the local pressure variations in the glottis. This pressure variations are caused by the flow of air which is expelled from the lungs through the bronchi and trachea by the muscle forces of the rib-cage (thorax), acting on the lung tanks. The flow of air out of the lungs is – at constant voice effort – an isobaric process since the lungs volume $V = V' + V''$ (Fig. 2a), represented in Fig. 2b by the condenser C_L, decreases as the quantity of air in the lungs diminishes, so that the relation of the charge Q_L to the capacity C_L (i.e. the voltage $U_L = Q_L/C_L$ on the equivalent condenser C_L) remains constant. It follows that the air pressure P_L in the lungs corresponding to the voltage U_L also remains constant, i.e. $P_L = \text{const}$. The larynx generator may thus be considered in future discussions and equivalent electric circuits as a system which is driven by voltage source of constant electromotive force $P_L = \text{const}$ with zero internal impedance. This has been shown in Fig. 2c, where the conductance G_L of the condenser C_L, representing the negligible acoustic loss resistance of the spongy and vesicular walls of the lung tanks has been disregarded.

The quasi-periodic vibrations of the vocal folds change the acoustic impedance $Z_g = R_g + j\omega L_g$ of the glottis modulating the air flow. At the entrance to the vocal tract the air flow has the form of recurrent discrete pulses. The latter play the rôle of an impulse source function, which excites vibrations of the air in the vocal tract at frequencies corresponding to the resonances

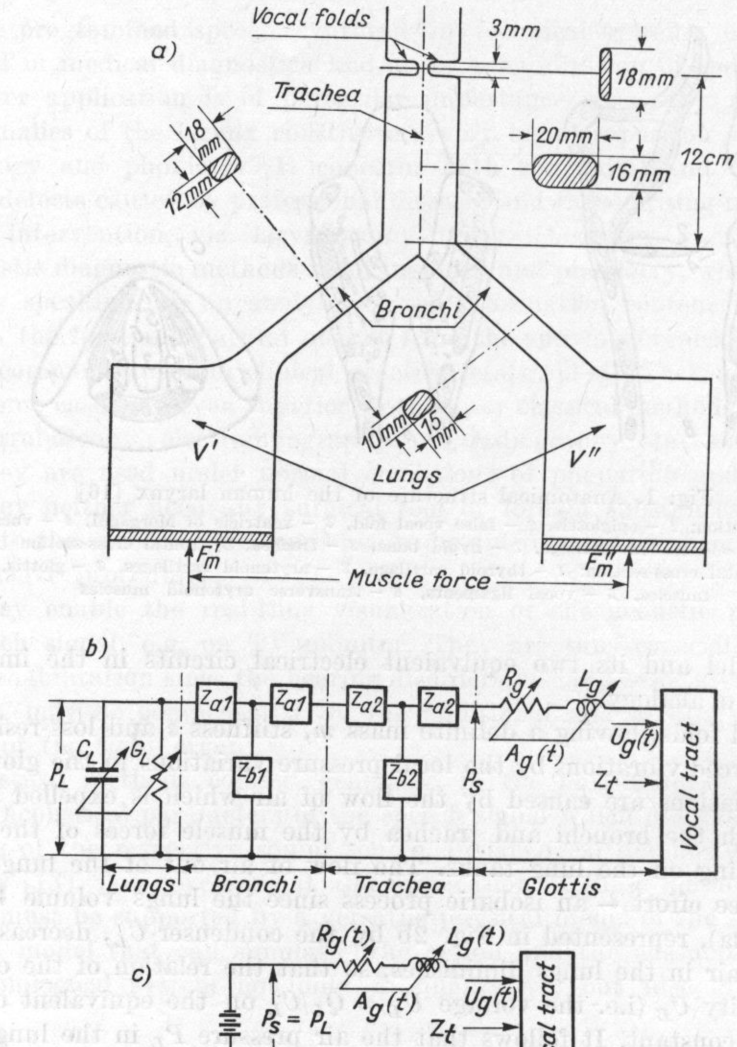

Fig. 2. Simplified physical model of the subglottal part of the human respiratory system (a) and its equivalent electrical circuits (b) (c)

of the vocal tract cavities. This resonances depend on the momentary geometrical configuration of the tract described by the tract's shape function $A_t(x)$.

In view of the relatively large cross-sectional areas (ca. 400 mm²) and small lengths (12 ÷ 15 cm) of the bronchial and tracheal tubes, represented in Fig. 2b in the form of the cascade connection of the four-poles T_1 and T_2, their longitudinal acoustic impedances are small and the corresponding pressure drops may be disregarded. Consequently, the subglottal pressure P_s is equal to the pressure P_L in the lungs, i.e. $P_s \approx P_L = \text{const}$. The total pressure drop

P_s appears across the glottis impedance Z_g and is transformed into the kinetic energy of the air flow, which is expressed in terms of the volume velocity U_g of the larynx pulses $U_g(t)$. This motivates the final simplification of the equivalent circuit in Fig. 2b to the form shown in Fig. 2c.

2.2 *Acoustic impedance of the glottis.* In Fig. 3 the glottis orifice is represented in the form of a parallelepiped slit whose geometrical dimensions correspond to the average anatomical glottis dimensions of an adult male. The effective glottis area $A_g(t) = lw(t)$ is a function of the width $w(t)$ which varies quasi-periodically as the vocal folds vibrate. In spite of such an extreme simplification for the glottis' model, its acoustic impedance $Z_g = R_g + j\omega L_g$ cannot be expressed analytically by the well known formula valid for a narrow slit, as has been done in the general theory of acoustic elements and circuits (cf. e.g. [17]). The pioneers of the larynx source theory, WEGEL [15] and van den BERG et al. [1], proved that the real component R_g of the glottis impedance is the sum of two components,

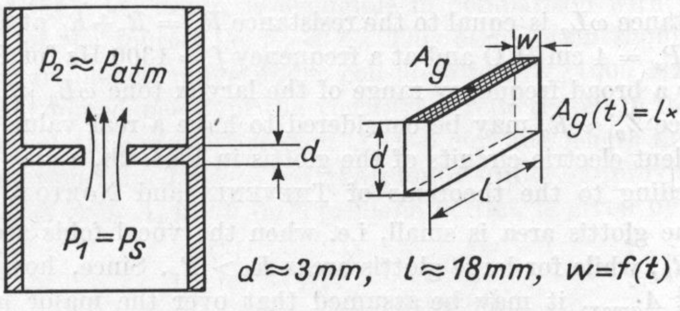

Fig. 3. Simplified physical model of the glottis

$$R_g = R_v + R_k = 12\mu d l^2 A_g(t)^{-3} + 0.44\varrho A_g(t)^{-2}\,|U_g(t)|, \tag{1}$$

where $R_v \doteq A_g(t)^{-3}$ is the classical viscous loss resistance of the medium with dynamical viscosity coefficient μ, flowing through a slit of effective area $A_g(t)$, length l and depth d, $R_k \doteq A_g(t)^{-2}\,|U_g(t)|$ is the kinetic resistance referred to the process of transforming the pressure drop P_s in the glottis into kinetic energy of medium flow, according to the formula

$$P_s = \tfrac{1}{2}\varrho u^2 = \tfrac{1}{2}\varrho(U_g/A_g)^2,$$

in which u denotes the particle velocity, U the volume flow velocity and ϱ is the density of the medium.

The previous model investigations have proved that formula (1) is valid for a broad range of flow parameters: $P_s \leqslant 64$ cm H_2O, $0.1 \leqslant w(t) \leqslant 2$ mm, $|U_g| \leqslant 2000$ cm³/s.

The reactance $j\omega L_g$ is the classical inertia of the air vibrating in the glottis and, ignoring edge-effects at the inlet and outlet of the orifice, may be expressed by the formula

$$j\omega L_g = j\omega \varrho d A_g(t)^{-1}, \tag{2}$$

where ϱ is the air density, d the depth of the slit, and $A_g(t)$ is its effective area perpendicular to the direction of flow.

Having calculated the numerical values of the components R_v, R_k and ωL_g of the glottis impedance $Z_g = (R_v + R_k) + j\omega L_g$ according to formulae (1) and (2), one can draw the following conclusions which are of use in the design of the larynx-source model:

(a) The reactance ωL_g is equal to the resistance $R_g = R_v + R_k$ at a frequency $f = 700$ Hz for $P_s = 4$ cm H_2O and at a frequency $f = 1300$ Hz for $P_s = 16$ cm H_2O. Thus over a broad frequency range of the larynx tone $\omega L_g \ll R_g$ and the glottis impedance $Z_g \approx R_g$ may be considered to have a real value. It follows that the equivalent electric circuits of the glottis in Figs. 2b, c may be further simplified according to the theorems of THEVENIN and NORTON.

(b) When the glottis area is small, i.e. when the vocal folds are near one another, $R_v > R_k$, while for large glottis areas $R_k > R_v$. Since, however, $R_v \approx \approx R_k$ at $A_g \approx \frac{1}{5} A_{g\max}$, it may be assumed that over the major part of the vibration cycle the glottis resistance is determined by its kinetic component and, consequently, $R_g \approx R_k$.

(c) The time constant $\tau = L_g/R_g$ is always smaller than both L_g/R_v and L_g/R_k and does not exceed the value $\tau_{\max} = 0.25$ ms (Fig. 4). The time constant $\tau = L_g/R_g$ is thus one order of magnitude smaller than the shortest period

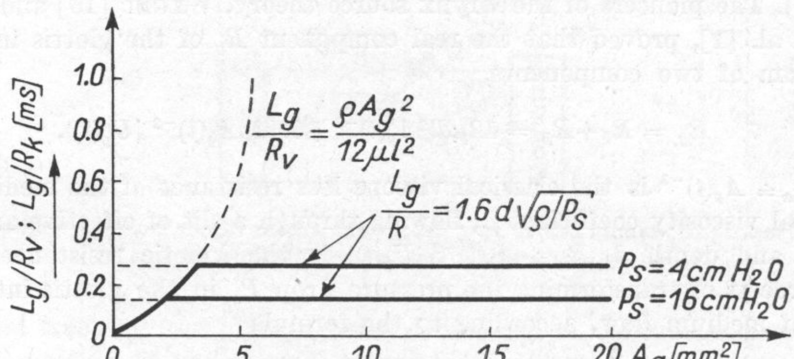

Fig. 4. Time constant L_g/R_v and L_g/R_k of the larynx oscillator as functions of the area A_g of the glottis orifice [4]

of vocal fold vibration $T_{\min} = 2.5$ ms, which corresponds to the highest larynx tone frequency of male voices $F_0 \approx 400$ Hz. It follows that the flow of air through the glottis, i.e. the volume velocity function U_g (t), may be determined from the formula

$$U_g(t) = P_s/R_g(t) \approx P_s/R_k(t), \tag{3}$$

which is valid for steady flow only.

Under average anatomical conditions of an adult male ($A_{gav} \approx 5$ mm^2), at medium voice effort ($P_s \approx 10$ cm H_2O) and over the frequency range $f \ll 5000$ Hz, the acoustic impedance of the glottis $Z_g \approx [90+j\omega 7\times 10^{-3}]10^5$ MKS acoustic ohms, and its modulus $|Z_g| \approx 100\cdot 10^5$ MKS acoustic ohms.

2.3. *Acoustic input impedance of the vocal tract.* Formula (3), which determines the volume velocity $U_g(t)$ of the glottal source, was a priori based on the assumption that the load impedance of the source, i.e. the input impedance Z_t of the vocal tract, is negligible in comparison with the internal impedance Z_g of the source, i.e. $Z_t \ll Z_g$. This assumption should now be justified for the frequency range under consideration $f \leqslant 1000$ Hz.

The input impedance Z_t of the vocal tract represented in the form of a lossless cylindrical tube of cross-section $A_t = 5$ cm^2 and length $l_t = 17$ cm, loaded by the radiation impedance Z_r of the mouth orifice, approximated by a circular piston of area A_t in an infinite plane baffle[1], is given by the formula

$$Z_t = Z_0\frac{Z_r+Z_0\tan(\beta l_t)}{Z_0+Z_r\tan(\beta l_t)}, \tag{4}$$

in which $Z_0 = \varrho c/A_t$ is the characteristic impedance of the tube, $\beta = \omega/c = 2\pi/\lambda$ is the propagation constant, Z_r is the radiation impedance of a circular piston of radius r and $A_t = \pi r^2$ is the area.

Hence

$$Z_r \approx \frac{\varrho c}{A_t}\left[\frac{(\beta r)^2}{2}+j\frac{8}{3\pi}(\beta r)\right]. \tag{5}$$

In model investigations the radiation impedance Z_r (5) may be represented in the form of the parallel connection of the radiation resistance R_R,

$$R_R = 128/9\pi^2\left(\frac{\varrho c}{A_t}\right), \tag{5a}$$

and the inductance L_R,

$$L_R = 8r/3\pi c\left(\frac{\varrho c}{A_t}\right), \tag{5b}$$

which simulates the mass of the co-vibrating medium (cf. e.g. [4], p. 33).

It may be proved [10] that, according to the general theory of acoustic wave-guides, the function $Z_t(f)$ (4) depends strongly on frequency, and its

[1] This is the commonly accepted physical simulation of articulation conditions in the case of the neutral mouth vowel Ǝ.

maximum values $Z_{t\max}$, which correspond to $\lambda/4$ – resonances of the vocal tract, that is to odd multiples of 500 Hz, decrease asymptotically with increasing frequency. The decreasing $Z_{t\max}$ values are comparable with the acoustic impedance Z_g of the glottis only in the vicinity of the first formant $F_1 \approx 500$ Hz ($Z_{t\max F1} \approx 86\times 10^5 \times e^{-j77°}$ MKS acoustic ohms) and of the second formant $F_2 \approx 1500$ Hz ($Z_{t\max F2} \approx 24\times 10^5 \times e^{-j56°}$ MKS acoustic ohms), approaching a limiting value equal to the characteristic impedance of the vocal tract itself $Z_0 = 8.5\times 10^5$ MKS acoustic ohms.

It thus follows that the a priori condition $Z_t \ll Z_g$ is fulfilled over the whole of the frequency range under consideration. The larynx source may thus be treated as a constant current source since it delivers a constant volume velocity, independent of the momentary configuration of the vocal tract which affects the value and ahape of the flow function in the vicinity of the first and second formants only. This influence may, in fact, be observed in experimental investigations.

2.4. *Time and frequency characteristics of the larynx source.* Typical examples of the laryngeal pulses are shown in Fig. 5. Each pulse is represented by a single period of the glottis area function $A_g(t)$ during phonation (continuous line) and by a period of the flow function, i.e. the volume velocity $U_g(t)$ in the glottis (broken line) [2]. Both functions correspond to the articulation of the neutral vowel Ǝ in various conditions of phonation ($P_s = 4 \div 24$ cm H_2O, $F_0 = 111 \div 250$ Hz). In view of the previously discussed independence of the source characteristics and the vocal tract configuration, the laryngeal pulses in Fig. 5 can also be considered to be representative for all other voiced sounds. The comparison of the functions $U_g(t)$ and $A_g(t)$ suggests the following conclusions:

(a) Both functions are triangular – a general and common feature of laryngeal pulses. The slopes of the flow pulses $U_g(t)$ are, however, somewhat steeper than those of the area pulses $A_g(t)$. This fact might be indirectly deduced from formula (3) in which the resistance $R_g(t)$, determined by the expression (1), is inversely proportional to the second or third power of $A_g(t)$. The steepness of the flow function pulses $U_g(t)$ results in somewhat higher levels of spectral components in the upper frequency range, as compared with those of the area function $A_g(t)$.

(b) An increase of the vocal effort or phonation intensity, which corresponds to higher values of the subglottal pressure P_s, results in a shortening of the open period τ_0 of the glottis. This effect may be expressed analytically by the decrease in the duty factor $\vartheta = \tau_0/T_0$, from $\vartheta \approx 1$ at $P_s = 4$ cm H_2O to $\vartheta = 0.5 \div 0.6$ at $P_s = 24$ cm H_2O, provided that the fundamental frequency F_0 of the larynx tone is kept constant.

The spectral characteristics of the larynx source function $U_g(t)$ or $A_g(t)$ are normally considered in the complex frequency plane $s = \sigma + j\omega$ using the

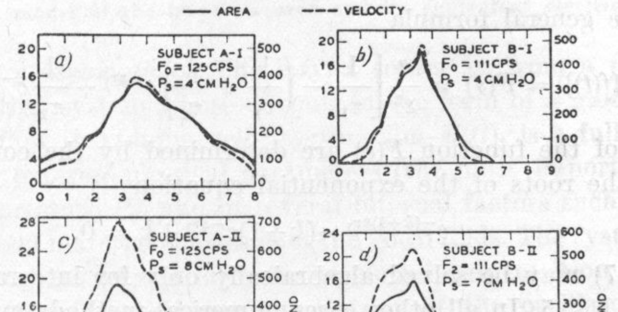

Fig. 5. Typical examples of the larynx pulses at various fundamental frequencies F_0 and different vocal efforts P_s [2]. The glottis area function $A_g(t)$ – continuous line. The flow function $U_g(t)$ – broken line

Laplace transformation. In Fig. 6 an approximation of the larynx pulse is shown in the form of a triangle of amplitude a, rise time τ_1, decay time $\tau_2 = k\tau_1$, length $\tau_0 = \tau_1 + \tau_2 = (k+1)\tau_1$ and duty factor $\vartheta = \tau_0/T_0$, where $T_0 = 1/F_0$ is the larynx tone period. The Laplace transformation of such a pulse may be ex-

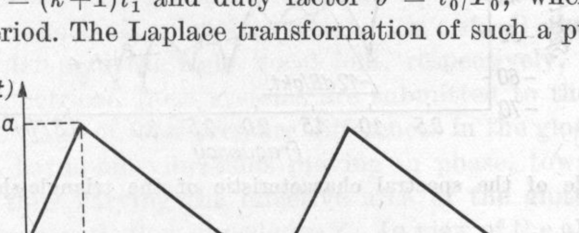

Fig. 6. Approximation of the larynx pulse in the form of a triangle

pressed by the general formula

$$\mathscr{L}\{f(t)\} = F(s) = \frac{a}{s^2}\left[\frac{1}{\tau_1} - \left(\frac{1}{\tau_1}+\frac{1}{\tau_2}\right)e^{-s\tau_1} + \frac{1}{\tau_2}e^{-s\tau_0}\right]. \tag{6}$$

The zeros of the function $F(s)$ are determined by the complex values of s, which are the roots of the exponential equation

$$e^{-(k+1)s\tau_1} - (k+1)e^{-s\tau_1} + k = 0. \tag{7}$$

Equation (7) may be solved algebraically only for integral values of k in the range $1 \leqslant k \leqslant 5$. In all other cases numerical methods must be used. Detailed consideration of the distribution of the zeros of function (6) in the complex frequency plane $s = \sigma + j\omega$ for different shapes of larynx pulses may be found in chapter 6.24 of reference [4]. In the particular though frequently-occurring case of a symmetrical larynx pulse, when $\tau_1 = \tau_2$ and $k = 1$, equation (7) reduces to the quadratic form [10]

$$e^{-2s\tau_1} - 2e^{-s\tau_1} + 1 = 0, \tag{8}$$

whose roots are given by

$$s_{2n} = \pm jn\frac{2\pi}{\tau_1} = \pm j2n\frac{2\pi}{\tau_0} \quad (n = 1, 2, 3, \ldots). \tag{9}$$

In this particular case the minima of the spectral envelope occur at the frequencies $f_{2n} = 2n/\tau_0 = 2/\tau_0;\ 4/\tau_0;\ 6/\tau_0$ etc., and the average level of the envelope, expressed by a function of the type $[(\sin nx)/nx]^2$ is inversely proportional to the square of the frequency and decreases at a rate of 12 dB per octave, see Fig. 7.

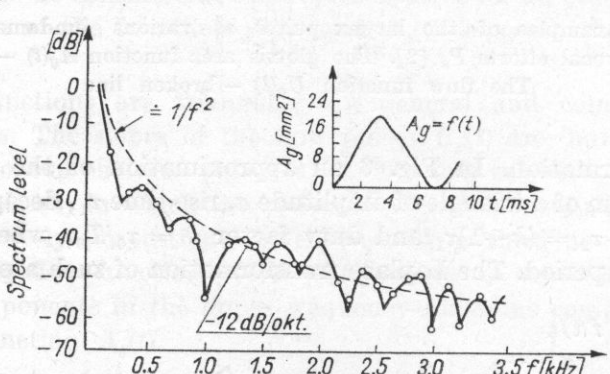

Fig. 7. Example of the spectral characteristic of the triangle-shaped larynx pulse

Since the Laplace transformation is a reversible operation, the spectral characteristic $F(s)$ of a voiced segment of the speech signal, previously submitted to the inverse filtering process [12], may be used for the analytical evaluation of the shape of the larynx pulses $U_g(t) = \mathscr{L}^{-1}\{U(s)\}$. This fact is very important in computer-aided phoniatric diagnostics.

3. Mechanical model of the larynx source and its equivalent electrical circuit

3.1. *Preliminary assumptions.* The larynx source is, from a physical point of view, a generating system whose output, in the form of a glottis area function $A_g(t)$ and a flow or volume velocity function $U_g(t)$, is a fully determined function of a few exterior physical parameters (the most important of which is the subglottal pressure P_s) and of several internal factors such as the structural parameters and material constants of the vocal folds. The system as a whole is subjected to feed-back in the form of the glottal area function $A_g(t)$ and the flow function $U_g(t)$, which – being the output of the oscillator – influence its internal parameters, i.e. those which determine the glottis impedance Z_g. These relations are illustrated by the block diagram shown in Fig. 8.

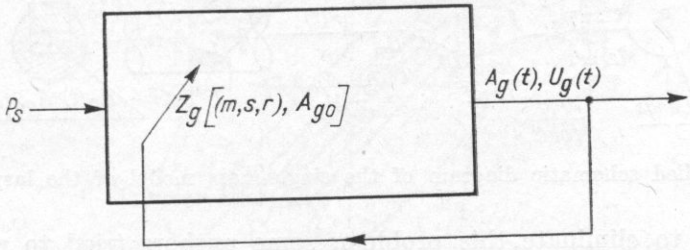

Fig. 8. Block diagram of the larynx generator as a feed – back system

A versatile physical model of the larynx source, adapted to acoustical diagnostic investigations, should enable us to determine the quantitative relations between the internal parameters of the source itself and its output $U_g(t)$ expressed in terms of directly measurable acoustic quantities, such as the acoustic pressure $p(t)$ of the sound wave in front of the patient's mouth.

3.2. *Mechanical model of the larynx source.* In the majority of papers concerning the larynx source, the vocal folds are presented in the form of a simple mechanical vibrating system with lumped constants, which consists of two indeformable masses m', m'', springs of stiffness s', s'' and viscous loss resistances r', r''. As a rule, both vibrating systems (m', s', r') and (m'', s'', r''), which simulate the left and the right vocal fold, respectively, are considered to be mutually symmetrical. Both systems are submitted to the action of the subglottal pressure P_s and of local pressure differences in the glottis. As a result they perform quasi-harmonic vibrations moving in phase, towards and away from one another, thus varying the effective area of the glottis orifice $A_g(t)$ and consequently its acoustic flow impedance Z_g. In view of the above symmetry of the vibrating systems, they may be simplified to the form of a single vibrating system with one mass, one degree of freedom, and equivalent parameters $m = m'+m''$, $s = s'+s''$, $r = r'+r''$ equal to the sums of the respective partial constants. The detailed analysis of such a single-mass model may be found in [4, 5]. The simplified scheme of the single-mass model of the larynx source is shown in Fig. 9, according to [4].

This model, though relatively simple and thus convenient for analytical interpretation, proved, however, to be unsatisfactory in phoniatric diagnostics. The reason for this was that it did not take into account an essential physiological factor connected with the anatomical structure and the cinematics of the vocal folds. This feature is the deformability of the vocal folds resulting in an asymmetry of vibration and in phase shifts between the displacements of their lower and upper parts or edges in quasi-harmonic motion.

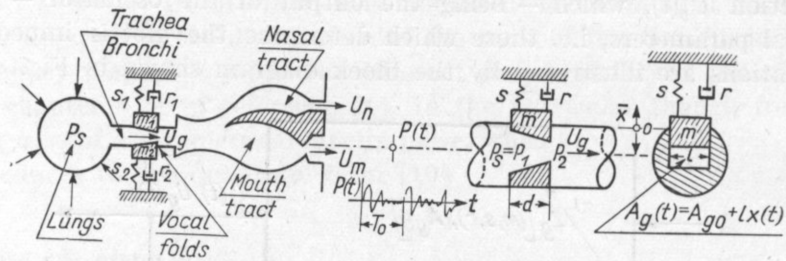

Fig. 9. Simplified schematic diagram of the single-mass model of the larynx source [4]

In order to eliminate this problem, some authors tried to represent the vocal folds as mechanical systems with continuously distributed constants, that is consisting of an infinite number of elementary masses, springs and loss resistances. Such a proposition [6], although theoretically based and quite possible in practice, due to the almost unlimited facilities offered by computing techniques, is neither necessary nor convenient in medical diagnostics, since the fundamental physiological and pathological features of the human larynx source may be simulated with sufficient accuracy in the two-mass model with lumped constants, proposed by Ishizaka and Matsuidara [7], presented by Flanagan [3] to the 7th ICA in Budapest and described in [9].

3.3. *The two-mass model of the larynx source.* Figure 10 presents a simplified scheme of the two-mass model in two cross-sectional planes: the vertical X-Y (a) and the horizontal X-Z (b), and its situation in the X-Y-Z axis system (c). The essential characteristic features of the two-mass model are the following:

(a) The depth d of the glottis slit is divided into two segments d_1 and d_2 ($d = d_1+d_2$), which correspond to the masses m_1 and m_2 representing the lower and the upper parts of the vocal fold, respectively. The areas of the glottis orifice in the segments d_1 and d_2 are given by the expressions

$$A_{g1} = A_{g01}+2x_1(t)l_g, \tag{10a}$$

$$A_{g2} = A_{g02}+2x_2(t)l_g, \tag{10b}$$

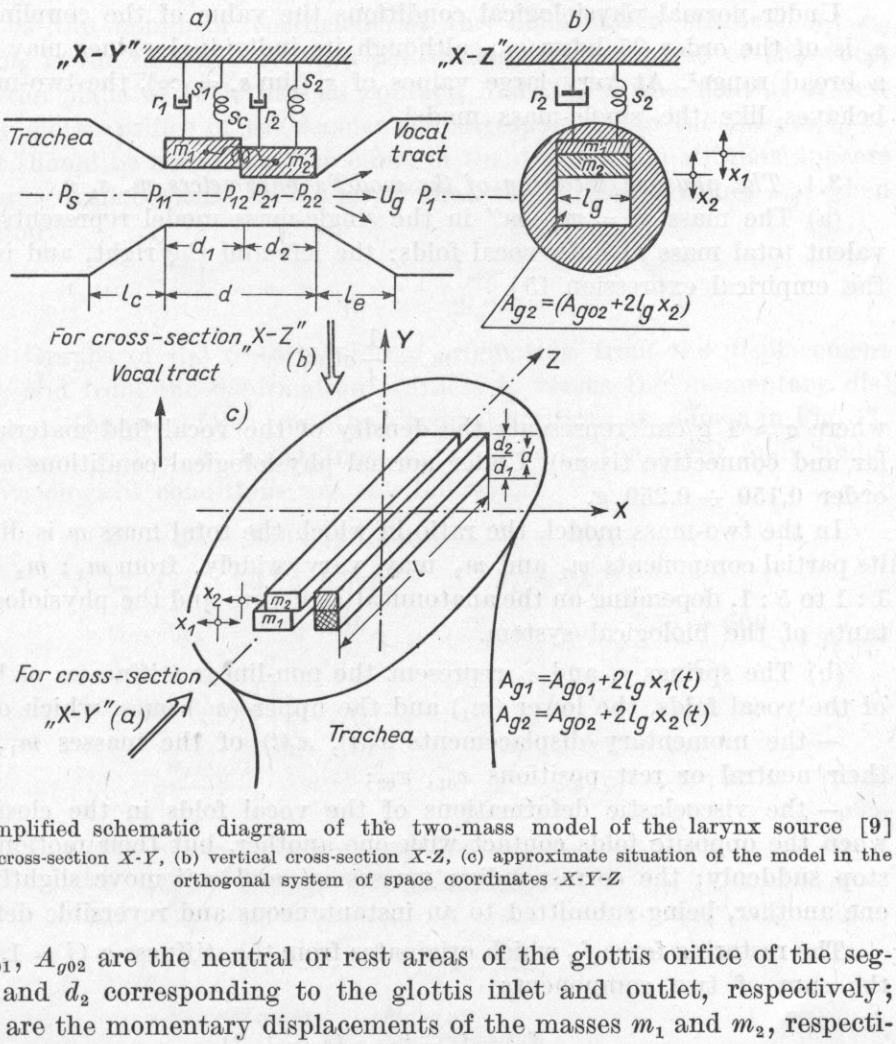

Fig. 10. Simplified schematic diagram of the two-mass model of the larynx source [9]
a) horizontal cross-section *X-Y*, (b) vertical cross-section *X-Z*, (c) approximate situation of the model in the orthogonal system of space coordinates *X-Y-Z*

where A_{g01}, A_{g02} are the neutral or rest areas of the glottis orifice of the segments d_1 and d_2 corresponding to the glottis inlet and outlet, respectively; $x_1(t)$, $x_2(t)$ are the momentary displacements of the masses m_1 and m_2, respectively, from the neutral or rest positions.

(b) The masses m_1 and m_2 are mutually coupled by the spring s_c which represents the bending stiffness of the vocal folds in the vertical plane *X-Y* parallel to the direction of vibration. The coupling stiffness s_c is, in fact, a non--linear parameter which, however, may be linearized without any essential influence on the accuracy of the model. The restoring force f_c originating from the coupling stiffness s_c is given by the expression

$$f_c = 2s_c(x_1 - x_2)^3(d_1 + d_2)^{-2}, \tag{11a}$$

which, provided the system is linear, may be rewritten in the following simplified form:

$$f_c \approx s_c(x_1 - x_2). \tag{11b}$$

Under normal physiological conditions the value of the coupling stiffness s_c is of the order 25 kdyn/cm, although its individual values may vary over a broad range[2]. At very large values of s_c ($\lim s_c = \infty$), the two-mass model behaves like the single-mass model.

3.4. *The physical meaning of the model's parameters m, s, r.*

(a) The mass $m = m' + m''$ in the single-mass model represents the equivalent total mass of both vocal folds: the left and the right, and is given by the empirical expression [5]

$$m = \frac{1}{4}\sigma l_g^2 d, \tag{12}$$

where $\sigma \approx 1$ g/cm³ represents the density of the vocal fold material (muscular and connective tissue). Under normal physiological conditions m is of the order $0.150 \div 0.250$ g.

In the two-mass model, the ratio in which the total mass m is divided into its partial components m_1 and m_2 may vary widely, from $m_1 : m_2 = d_1 : d = 1:1$ to $5:1$, depending on the anatomical structure and the physiological constants of the biological system.

(b) The springs s_1 and s_2 represent the non-linear stiffnesses of both parts of the vocal folds, the lower (m_1) and the upper (m_2) ones, which oppose:

– the momentary displacements $x_1(t)$, $x_2(t)$ of the masses m_1, m_2 from their neutral or rest positions x_{01}, x_{02};

– the viscoelastic deformations of the vocal folds in the closing phase, when the opposite folds contact with one another, but their motion does not stop suddenly; the corresponding masses m' and m'' move slightly towards one another, being submitted to an instantaneous and reversible deformation.

The restoring force f_{sj} which originates from the stiffness s_j ($j = 1, 2$) is thus the sum of two components

$$f_{sj} = f_{kj} + f_{hj} \quad (j = 1, 2). \tag{13}$$

The first component f_{kj} denotes the restoring force due to the displacement stiffness k_j and may be expressed by the formula

$$f_{kj} = k_j x_j (1 + \eta_{kj} x_j^2) \quad (j = 1, 2), \tag{14}$$

where η_{kj} is the non-linear coefficient of the displacement stiffness k_j.

The second component f_{hj} denotes the restoring force due to the deformation stiffness h_j and may by expressed by the formula

$$f_{hj} = h_j (x_j - x_{0j})[1 + \eta_{hj}(x_j - x_{0j})^2] \quad (j = 1, 2;\ x_j \geqslant x_{0j}), \tag{15}$$

[2] The numerical values of the equivalent mechanical constants of the biological larynx source system are quoted from [8].

where η_{hj} is the nonlinear coefficient of the deformation stiffness h_j, $x_{0j} = -A_{g0j}/2l_g$ is the distance between the neutral or rest plane of the vocal folds and the plane of their mutual contact, and A_{g0j} is the neutral or rest area of the glottis orifice of the segment d_j corresponding to the mass m_j ($j = 1, 2$). It should be noted that the effect of the deformation stiffness appears only for large displacements x_j, i.e. larger than the threshold values x_{0j}, when the condition

$$x_j \geqslant x_{0j} = -\frac{A_{g0j}}{2l_g} \tag{16}$$

is fulfilled. Graphs of the restoring forces originating from the displacement stiffness k_j and from the deformation stiffness h_j versus the momentary displacement x_j of the vocal folds from their neutral positions are shown in Fig. 11.

The average values of the stiffness parameters of the vocal folds under normal physiological conditions are the following:

$$k_1 = 50 \div 100 \text{ kdyn/cm}, \qquad h_1 \approx 3k_1,$$
$$k_2 = 5 \div 50 \text{ kdyn/cm}, \qquad h_2 \approx 3k_2,$$
$$\eta_{k1},\ \eta_{k2} = 50 \div 100, \qquad \eta_{h1},\ \eta_{h2} = 250 \div 500.$$

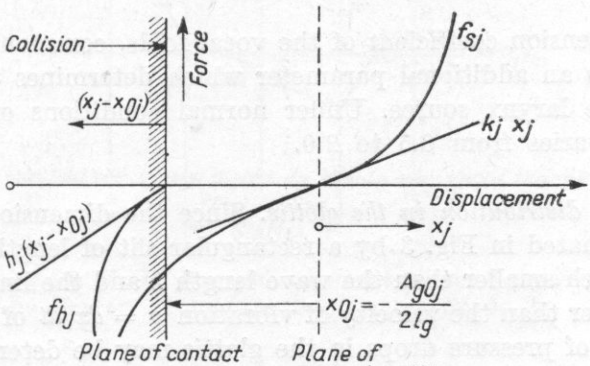

Fig. 11. Restoring forces originating from the displacement stiffness k_j and deformation stiffness h_j as functions of the momentary displacements x_j of the vocal folds from their rest positions

(c) The loss resistance r is attributed to:

– the viscous losses in the material, i.e. the muscular and connective tissue of the vocal folds,

– the mutual adhesion of the vocal folds contacting one another in the closing phase, when they move together and away from one another.

The influence of the loss resistance is an increase in the attenuation or damping coefficients ξ_1 and ξ_2 expressed by the formulae

$$\xi_1 = r_1/2\sqrt{m_1 k_1}, \qquad \xi_2 = r_2/2\sqrt{m_2 k_2}, \tag{17a, b}$$

in which k_1, k_2 are the linear coefficients of the displacement stiffness, cf. equation (14).

The average values of the damping coefficients ξ_1 and ξ_2 under normal physiological conditions are as follows:

– in the closing phase of the glottis: $\xi_1 \approx 0.1$, $\xi_2 \approx 0.6$;

– in the opening phase of the glottis: $\xi_1 \approx 1.1$, $\xi_2 \approx 1.6$.

(d) The tension coefficient Q of the vocal folds. The mass and stiffness parameters of the vocal folds determine the natural frequency f_0 of their oscillations, expressed by the formula

$$f_0 = \frac{1}{2\pi}\sqrt{\frac{s}{m}}. \tag{18a}$$

Under real conditions of phonation and articulation, due to the contraction of the vocalis muscles the vocal folds are shortened or elongated and their tension varies to a limited extent. Their mass m and stiffness s also vary and the frequency of oscillations changes from the value f_0 (18a) to the value f_0' given by the expression

$$f_0' = \frac{1}{2\pi}\sqrt{\frac{sQ}{m/Q}} = \frac{Q}{2\pi}\sqrt{\frac{s}{m}} = Qf_0, \tag{18b}$$

where Q is the tension coefficient of the vocal folds, commonly used in model investigations as an additional parameter which determines the real physical constants of the larynx source. Under normal conditions of phonation and articulation Q varies from 0.5 to 2.0.

3.5. *Pressure distribution in the glottis.* Since the dimensions of the glottal orifice, approximated in Fig. 3 by a rectangular slit of length l, width w and depth d, are much smaller than the wave length λ and the linear flow velocity u_g is much smaller than the velocity of vibration $v = dx/dt$ of the vocal fo ds, the distribution of pressure drops in the glottis may be determined according to the general rules for steady-state unidimensional flow of air along the Y axis, (see Fig. 10). The scheme of the longitudinal cross-section of the glottis model and the corresponding pressure drops ΔP along the flow axis are shown in Fig. 12a and 12b, respectively. The pressure drops are determined by the expressions (19) ÷ (23):

$$P_s - P_{11} = 0.69\varrho(U_g^2 A_{g1}^{-2}) + \int_0^{l_c} \frac{\varrho}{A_c(x)}\, dx \frac{dU_g}{dt}, \tag{19}$$

$$P_{11} - P_{12} = 12\mu d_1 l_g^2 A_{g1}^{-3} U_g + \frac{\varrho d_1}{A_{g1}} \frac{dU_g}{dt}, \tag{20}$$

$$P_{12} - P_{21} = \frac{\varrho}{2} U_g^2 (A_{g2}^{-2} - A_{g1}^{-2}), \tag{21}$$

$$P_{21}-P_{22} = 12\mu d_2 l_g^2 A_{g2}^{-3} U_g + \frac{\varrho d_2}{A_{g2}} \frac{dU_g}{dt}, \tag{22}$$

$$P_{22}-P_1 = -\varrho U_g^2 A_{g2}^{-2} \frac{A_{g2}}{A_t}\left(1-\frac{A_{g2}}{A_t}\right). \tag{23}$$

Formulae (19) - (23) are based on the following assumptions (see Fig. 12):

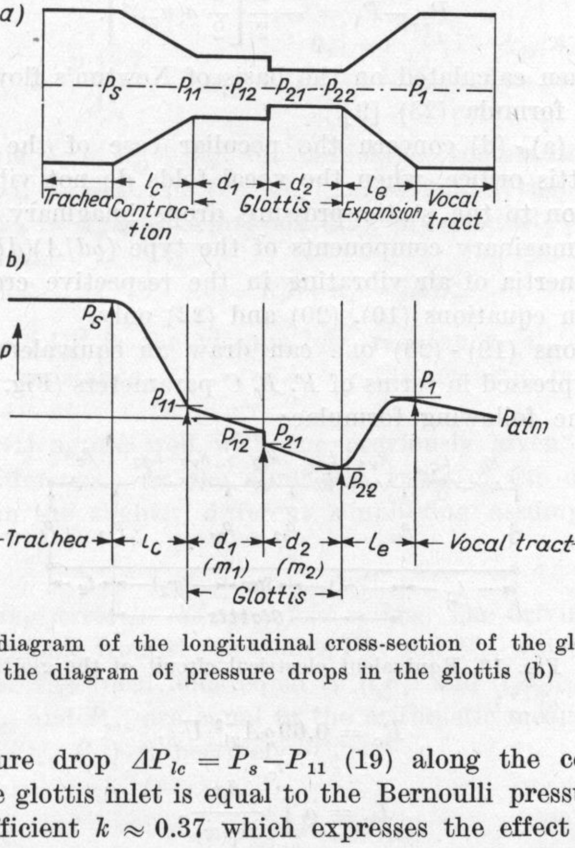

Fig. 12. Schematic diagram of the longitudinal cross-section of the glottis model (a) and the diagram of pressure drops in the glottis (b)

(a) The pressure drop $\Delta P_{lc} = P_s - P_{11}$ (19) along the constriction l_c of the trachea at the glottis inlet is equal to the Bernoulli pressure increased by an empirical coefficient $k \approx 0.37$ which expresses the effect of the constriction [1], so that, finally,

$$\Delta P_{lc} \approx 1.37 \frac{\varrho}{2} (u_{g1})^2.$$

(b) The pressure drops $\Delta P_{d1} = P_{11} - P_{12}$ (20) and $\Delta P_{d2} = P_{21} - P_{22}$ (22) along the segments d_1 and d_2, respectively, originate from the viscous loss resistances R_{v1} and R_{v2} determined by the general formula (1).

(c) The pressure drop $\Delta P_{d1/d2} = P_{12} - P_{21}$ (21) at the abrupt change of the glottis cross-sectional area is equivalent to the difference of the kinetic energy densities in the cross-sections A_{g1} and A_{g2}, provided that the air flow in the glottis is continuous: $U_g = \text{const}$. Hence

$$P_{12}-P_{21} = \frac{1}{2}\varrho U_g^2(A_{g2}^{-2}-A_{g1}^{-2}),$$

since $u_{g1}^2 = U_g^2 A_{g1}^{-2}$ and $u_{g2}^2 = U_g^2 A_{g2}^{-2}$.

(d) The pressure drop $\Delta P_{l_e} = P_{22}-P_1 \approx P_{22}-P_{\text{atm}}$ has a negative value; according to [1]

$$P_{22}-P_1 = -\frac{1}{2}\left[\frac{1}{2}\varrho(u_{g2})^2\right].$$

However, when calculated on the basis of Newton's flow laws, it has the value given by formula (23) [9].

Assumptions (a) - (d) concern the peculiar case of the static flow of air through the glottis orifice, when the vocal folds do not vibrate. In dynamic states of vibration to the static pressure drops imaginary components must be added. The imaginary components of the type $(\varrho d/A)(dU_g/dt)$, which are referred to the inertia of air vibrating in the respective cross-sections of the glottis, appear in equations (19), (20) and (22) only.

Using equations (19) - (23) one can draw an equivalent electrical circuit of the glottis, expressed in terms of R, L, C parameters (Fig. 13), whose values are given by the following formulae:

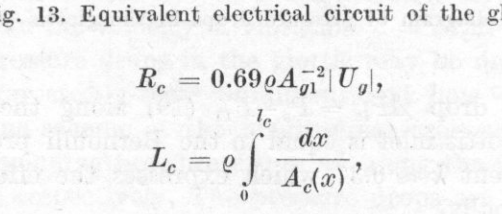

Fig. 13. Equivalent electrical circuit of the glottis

$$R_c = 0.69\varrho A_{g1}^{-2}|U_g|, \tag{24}$$

$$L_c = \varrho\int_0^{l_c}\frac{dx}{A_c(x)}, \tag{25}$$

$$R_{v1} = 12\mu l_g^2 d_1 A_{g1}^{-3}, \tag{26}$$

$$L_{g1} = \varrho d_1 A_{g1}^{-1}, \tag{27}$$

$$R_{v2} = 12\mu l_g^2 d_2 A_{g2}^{-3}, \tag{28}$$

$$L_{g2} = \varrho d_2 A_{g2}^{-1}, \tag{29}$$

$$R_{12} = \frac{1}{2}\varrho(A_{g2}^{-2}-A_{g1}^{-2})|U_g|, \tag{30}$$

$$R_e = -\varrho\frac{|U_g|}{A_{g2}A_t}\left(1-\frac{A_{g2}}{A_t}\right). \tag{31}$$

The acoustic impedance Z_g of the glottis in the two-mass model, determined on the basis of the equivalent circuit in Fig. 13, may be expressed as

$$Z_g = \frac{\varrho}{2}|U_g|\left[\frac{0.37}{A_{g1}^2}+\frac{1-2\frac{A_{g2}}{A_t}\left(1-\frac{A_{g2}}{A_t}\right)}{A_{g2}^2}\right]+(R_{v1}+R_{v2})+j\omega(L_{g1}+L_{g2}+L_c)$$

$$= (R_{k1}+R_{k2})|U_g|+(R_{v1}+R_{v2})+j\omega(L_{g1}+L_{g2}+L_c), \tag{32}$$

where

$$R_{k1} = \frac{0.19\varrho}{A_{g1}^2}, \quad R_{k2} = \varrho\frac{0.5-\frac{A_{g2}}{A_t}(1-A_{g2}/A_t)}{A_{g2}^2}. \tag{33a,b}$$

Since as a rule $L_c \ll L_{g1}+L_{g2}$, the inductance representing the inertia L_c may be neglected, $L_c \approx 0$. In the single-mass model the coupling stiffness $s_c = \infty$, $A_{g1} = A_{g2} = A_g$ and expression (32), provided $A_g \gg A_t$, may be rewritten in the form

$$Z_g' = R_v'+R_k'+j\omega L_g', \tag{34}$$

where $R_v' = 12\mu d l_g^2 A_g^{-3}$ is the viscous loss resistance, $R_k' = 0.69\varrho A_g^{-2}|U_g|$ – the kinetic flow resistance, $L_g' = \varrho d A_g^{-1}$ – the acoustic mass of the air in the glottis orifice.

Expression (34) agrees well with the previously given formulae (1) and (2). The only difference – in the numerical value of the constant factor in R_k' – results from the slightly different simplifying assumptions in the case considered.

3.6. *The driving forces in the vocal fold system.* The driving forces, denoted as F_1 and F_2, originate from the pressures P_{m1} and P_{m2} which act on the effective side areas of the vocal folds equal to $(l_g d_1)$ and $(l_g d_2)$, respectively. It is assumed that P_{m1} and P_{m2} are equal to the arithmetic means of the pressures (P_{11}, P_{12}) and (P_{21}, P_{22}) respectively, thus

$$P_{m1} = \frac{1}{2}(P_{11}+P_{12}), \tag{35}$$

$$P_{m2} = \frac{1}{2}(P_{21}+P_{22}). \tag{36}$$

Using equations (19)-(23), one can write

$$P_{m1} = \frac{1}{2}(P_{11}+P_{12}) = P_s-R_c U_g-\frac{1}{2}\left(R_{v1}U_g+L_{g1}\frac{dU_g}{dt}\right), \tag{35a}$$

$$P_{m2} = \frac{1}{2}(P_{21}+P_{22}) = P_{m1}-R_{12}U_g-\frac{1}{2}\left[(R_{v1}+R_{v2})U_g+(L_{g1}+L_{g2})\frac{dU_g}{dt}\right], \tag{36a}$$

and finally obtain

$$F_1 = P_{m1}(l_g d_1), \tag{37a}$$

$$F_2 = P_{m2}(l_g d_2). \tag{37b}$$

4. Electrical model of the larynx source and its mathematical description

4.1. *Electrical equivalent circuit of the model.* Having introduced the simplifying assumptions previously discussed which consist in:

(a) disregarding the acoustic impedances of the bronchi and trachea,

(b) representing the energy source, that is the lungs, by a d.c. voltage source with electromotive force $P_s \approx P_L$ and zero internal impedance,

(c) expressing the glottis impedance Z_g in the form given by equation (32), one may draw the electric equivalent circuit of the speech organ in vowel articulation as shown in Fig. 14. The pharynx-mouth vocal tract is simulated by the cascade connection of n symmetrical T-type four-poles, each of which represents an elementary section of the vocal tract approximated in the form of a cylindrical tube with hard walls, of cross-sectional area A_i and length l_i ($i = 1, 2, \ldots, n$). The acoustic parameters of the i-th four-pole T_i are as follows [10]:

$L_i = \varrho l_i / A_i$ – acoustic mass,

$C_i = A_i l_i / \varrho c^2$ – acoustic compliance,

$R_i = (S_i l_i / A_i^2)\sqrt{\varrho\mu\omega/2}$ – acoustic resistance of the viscous losses at the tube walls, where ϱ is the density of air, μ – the coefficient of viscosity, S_i – the tube circumference and $\omega = 2\pi f$ – the angular frequency. The neglected losses, caused by energy absorption at the walls of the tube and by heat conduction at the walls, may be taken into account by multiplying the acoustic loss resistance R_i, calculated at the frequency $f = F_0$, by the correction factor $k = 20$-25.

In the equivalent circuit given in Fig. 14, the radiation impedance Z_r is represented in the form of the parallel connection of the inductance L_R and radiation resistance R_R, according to the expressions (5), (5a) and (5b) in section 2.3.

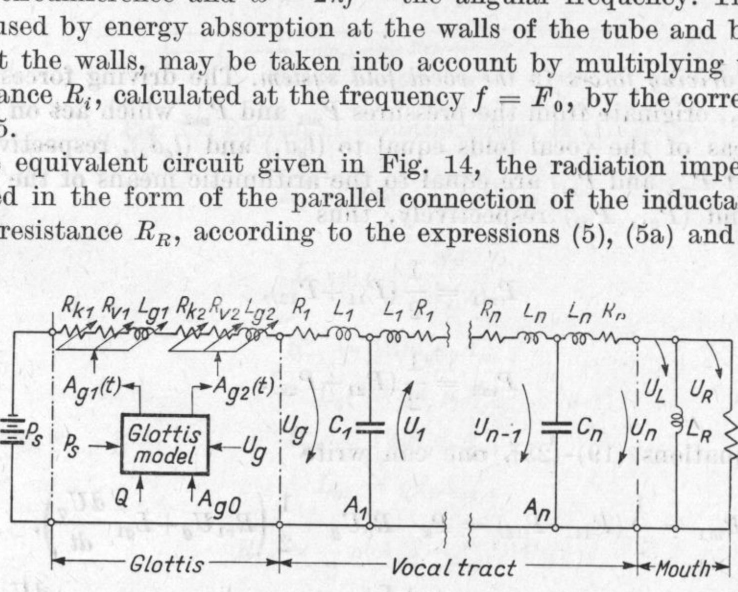

Fig. 14. Equivalent electrical circuit of the speech organ in vowel articulation

4.2. *Mathematical description of the model.* The behaviour of the electric circuit in Fig. 14 may be easily discussed by the classical methods of network analysis, e.g. by the loop-current method for $n+2$ loops, where n is the arbitrary number of elementary segments of the vocal tract. Let us for example put $n = 4$, thus simulating the vocal tract by four cylindrical segments only. In this case we obtain the set of six differential equations of motion (38)-(43):

$$(R_{k1}+R_{k2})|U_g|U_g+(R_{v1}+R_{v2})U_g+(L_{g1}+L_{g2})\frac{dU_g}{dt}+$$
$$+R_1U_g+L_1\frac{dU_g}{dt}+\frac{1}{C_1}\int_0^t(U_g-U_1)dt = P_s, \quad (38)$$

$$(R_1+R_2)U_1+(L_1+L_2)\frac{dU_1}{dt}+\frac{1}{C_2}\int_0^t(U_1-U_2)dt+\frac{1}{C_1}\int_0^t(U_1-U_g)dt = 0, \quad (39)$$

$$(R_2+R_3)U_2+(L_2+L_3)\frac{dU_2}{dt}+\frac{1}{C_3}\int_0^t(U_2-U_3)dt+\frac{1}{C_2}\int_0^t(U_2-U_1)dt = 0, \quad (40)$$

$$(R_3+R_4)U_3+(L_3+L_4)\frac{dU_3}{dt}+\frac{1}{C_4}\int_0^t(U_3-U_L)dt+\frac{1}{C_3}\int_0^t(U_3-U_2)dt = 0, \quad (41)$$

$$R_4U_L+(L_4+L_R)\frac{dU_L}{dt}-L_R\frac{dU_R}{dt}+\frac{1}{C_4}\int_0^t(U_L-U_3)dt = 0, \quad (42)$$

$$R_RU_R+L_R\frac{d}{dt}(U_R-U_L) = 0. \quad (43)$$

From these equations the currents (or volume velocities U) and voltages (or acoustic pressures p) in any loop and node of the network corresponding to a particular acoustic cross-section of the vocal tract may be determined. In acoustic diagnostics of the larynx source the following quantities and the mutual relationships are the most interesting:

(a) The volume velocity U_g in the glottis orifice as the output of the larynx source in the form of the flow function $U_g(t)$.

(b) The volume velocity U_R through the radiation resistance R_R of the mouth orifice, since this quantity is functionally connected to a directly measurable diagnostic parameter of the speech signal, viz. the acoustic pressure $p(r)$ at a distance r from the mouth orifice

$$p(r) = \frac{j\omega\varrho U_R}{4\pi r}e^{-j2\pi r/\lambda}. \quad (44)$$

Relation (44) is valid in the near field of a spherical wave, when $2\pi r \leqslant \lambda$.

The coefficients R_{k1}, R_{k2}, R_{v1}, R_{v2}, L_{g1}, L_{g2} in equation (38), which also appear in expression (32) for the acoustic impedance Z_g of the glottis, are functions of the glottis areas $A_{g1}(t)$ (10a) and $A_{g2}(t)$ (10b). This fact is attributed to the feed-back in the larynx generator, previously described in section 3.1. In this connection the flow equations (38)-(43) must be complemented by the differential equations of motion of the vocal folds, simulated by the masses m_1 and m_2. The equations of motion may be written in the form

$$F_1 = m_1 \frac{d^2x_1}{dt^2} + r_1 \frac{dx_1}{dt} + s_1 x_1 + s_c(x_1 - x_2), \tag{45a}$$

$$F_2 = m_2 \frac{d^2x_2}{dt^2} + r_2 \frac{dx_2}{dt} + s_2 x_2 + s_c(x_2 - x_1), \tag{45b}$$

where F_1 and F_2 are the driving forces given by formulae (35), (36) and (37).

For the purposes of numerical calculations and computer modelling of the speech organ in various conditions of phonation and articulation, the differential equations of flow (38)-(43) and motion (45a, b) should be made discrete by substituting for the differentiation $[df(t)/dt]$ and integration $[\int f(t)dt]$ operators the corresponding incremental approximations according to the rules

$$\frac{df(t)}{dt} \approx \frac{f(t_i) - f(t_{i-1})}{t_i - t_{i-1}} = \frac{f_i - f_{i-1}}{\tau}, \tag{46a}$$

$$\int f(t)dt \approx (t_i - t_{i-1}) \sum_{i=0}^{i-1} f_i = \tau \sum_{i=0}^{i-1} f_i, \tag{46b}$$

where τ is the step of time quantization, and i is the number of the successive time segments.

The flow and motion equations must be solved by an iterative method. The volume velocities U in the different cross—sections of the vocal tract, the driving forces F_1, F_2 and the resulting displacements x_1, x_2 of the vocal folds m_1, m_2 in successive i-th time segments are determined from the values of x_1 and x_2, calculated or assumed for the preceeding, that is (i-1)-th, time segment.

5. Conclusions

The preceeding theoretical considerations indicate the applicability and usefulness of a physical model of the human larynx source in diagnostic investigations of the speech organ in both normal and pathological cases. Any anomaly of the anatomical structure of the larynx and restriction of its motive capability, resulting either from laryngeal nerve palsy or from necessary surgical intervention, may be expressed in terms of the physical parameters of the model, which at the same time result in measurable variations of the acoustic structure of the speech signal being analysed in the time and/or frequency domains. These relations are mutual and reversible: the analysis of the acoustic parameters of the speech signal under definite phonation and articulation conditions makes it possible to determine, when using the model and its mathematical description, the equivalent values of the physical parameters of the larynx model and hence to define and localize any anomaly of the anatomical structure or reason for a restricted motive capability of the biological system. This in turn facilitates and makes objective the process of diagnosis.

The directly measurable acoustical diagnostic signal usually used in laryngology and phoniatry is the acoustic pressure $p(t)$ of the speech wave at a particular point on the axis of symmetry of the patient's mouth, or the spectral presentation of this function obtained by a Fourier or Laplace transformation. The acoustic pressure is connected with the volume velocity U_R in the mouth orifice by the simple relation (44). The time and frequency parameters of the $U_R(t)$ function contain essential information concerning the physical and material constants of the larynx source, as well as the spectral (e.g. formant-antiformant) structure of the speech signal-which depends on the momentary geometrical configuration of the vocal tract. The latter information may, if necessary, be eliminated by the process of inverse filtering. In this way the information content of the speech signal may be limited to the larynx source characteristics only. The essential diagnostic information is contained in the volume velocity function $U_g(t)$ through the glottis, since this function is related by the flow equations (38)-(43) and the motion equations (45) with the most important physical and physiological parameters of the vocal folds and with the subglottal pressure P_s.

The physical interpretation of the larynx pathology and the description of anatomical anomalies in terms of the physical parameters of the diagnostic speech signal is the subject of further research work which is being carried out in the Speech Acoustics Laboratory, Department of Cybernetic Acoustics, IFTR[3]) – Polish Academy of Sciences, in close co-operation with the Phoniatric Centre[4]) of the Otolaryngological Clinic, Central Clinical Hospital, Medical Academy in Warsaw.

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