An Impedance Detector for Glottography

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ABSTRACT

Impedance glottography, or electroglottography, is a noninvasive technique for monitoring the variation of the degree of contact between the vibrating vocal cords during voice production. Electrical impedance is sensed using a pair of electrodes placed on either side of the larynx, by injecting a low-level high-frequency current. A wide-band impedance detector circuit has been developed using a precision rectifier based on voltage feedback clamping amplifier and approximately linear-phase lowpass filter. The circuit can be used for obtaining electroglottogram signal for diagnosis of vocal fold disorders.

Keywords:
Bioimpedance sensor, Electroglottography, Glottography.

1. INTRODUCTION

Inside the larynx, there are two folds of muscular bundle, known as vocal cords, with a variable opening between them known as the glottis. These folds are set in vibration during the production of voiced speech segments. The vibration consists of three phases, namely, closure phase, separation phase and open phase [1,2]. The rate of vibration is known as the fundamental frequency, or the pitch, (F_o). The average pitch varies with the size of the speaker's vocal folds over a range of 80 to 250 Hz for male speakers and 200 to 450 Hz for female speakers. Monitoring the pattern of vocal fold vibrations is important in diagnosing laryngeal diseases.

Impedance glottography, also known as electroglottography or laryngography, is a noninvasive low-cost technique for monitoring vibration pattern of vocal folds, particularly variation in the degree of contact between them during the closure phase. It provides an electrical signal, known as electroglottogram (EGG) or laryngogram (Lx), proportional to variation in electrical impedance across a pair of disc electrodes placed in contact with the skin on either side of the thyroid cartilage of the larynx [2-10].

This paper describes a wide-band impedance detector circuit using a precision rectifier based on voltage feedback clamping amplifier and approximately linear-phase lowpass filter. The circuit can be used for obtaining electroglottogram signal for speech research and diagnosis of vocal fold disorders.

2. IMPEDANCE GLOTTOGRAPHY

The technique has been in use for about five decades, and clinical studies using several instruments (F-J Electronics, Denmark; Laryngograph, UK; Synchrovoice Research, USA; etc.) have been reported. A relationship between a typical EGG waveform and various vibration phases is shown in Figure 1, with four important points I1-I4 marked on the waveform. The first point I1 corresponds to the time when the vocal folds begin to open. Point I2 indicates the maximum opening. The third point indicates beginning of the closure and the last point indicates total closure. Variation in the location of these points may be indicative of vocal fold disorders [2,4,9,10]. A number of features of the waveform during sustained vowels, such as speed quotient, open quotient, closed quotient and pitch period, etc.; and multiple-period histograms of pitch periods obtained over sentences and passages have been used in clinical studies.

An impedance glottograph consists of an impedance sensor and an amplitude-modulation (AM) detector, as shown in Figure 2. Impedance sensing may be carried out by passing a high-frequency (100 kHz – 1 MHz) and low-level (< 10 mA) current through a pair of electrodes placed on either side of the larynx. The impedance variation caused by varying contact area between the vocal folds results in amplitude-modulated voltage across the

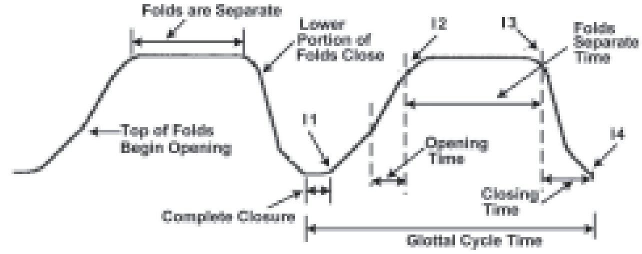

Figure 1: Relationship between EGG waveform and vibration phases of vocal folds (adapted from [10]).

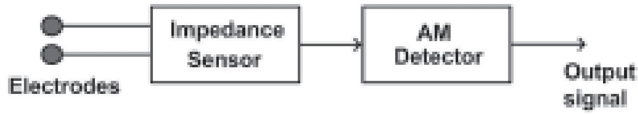

Figure 2: Block diagram of an impedance glottograph.

electrodes. This voltage is demodulated for obtaining an output related to the variation in the impedance. In another method, a voltage is applied across the electrode pair and the resulting current is sensed by a current-to-voltage converter, and the output is demodulated. In both the techniques, high-frequency transformers are used to isolate the body tissue from sensing circuit and the signal-acquisition system [4,6,11,12].

An excitation frequency in the range of 100 kHz to 1 MHz is used to ensure that the impedance is purely resistive, and variations in the impedance are contributed by glottal contact phases. At higher frequencies, the current remains confined to the surface and modulation index decreases, resulting in low sensitivity. At lower frequencies, the impedance may not remain purely resistive, and rejection of the carrier in the demodulator becomes difficult. The basal impedance of larynx region is normally in the range of 100 to 500 Ω, and the variation caused by vibration of the vocal cords is typically less than 1 Ω. For faithful sensing of impedance variation, the overall signal-to-noise ratio should be higher than 40 dB [4,6]. As the EGG signal may have spectral components in the band of 75 Hz to 8 kHz, the lowpass filter of the detector should have nearly flat magnitude and approximately linear-phase response over this band.

In the techniques reported by Fourcin [11] and Rothenberg [12], high-frequency voltage is applied for sensing the variation in the glottal impedance. A schematic representation of Fourcin's impedance sensor is shown in Figure 3a. It uses two disc electrodes with guard rings which are shorted together to reduce external interference [4,11]. It uses two isolation transformers. In the figure, the impedance between the central discs is modeled as a base resistance *Ro* in series with a small time-varying component $r(t)$. The output voltage for this sensor circuit is

$$v_0 \approx \frac{\beta}{\alpha} v_s \left(\frac{R_x/\beta^2}{R_0' + R_x/\beta^2} \right) \left(1 - \frac{r}{R_0' + R_x/\beta^2} \right) \quad (1)$$

where $R_0' = R_0 + R_s/\alpha^2$. Figure 3b and 3c show the two possible electrode configurations for a single channel of a multi-channel technique reported by Rothenberg [12]. Both use disc electrodes without guard rings. In Figure 3b, the electrodes are connected in parallel to the excitation and sensing circuits, with the output given as

$$v_o \approx \frac{\beta}{\alpha} v_s \frac{R_0}{R_0' + R_s/\alpha^2} \left(1 + \gamma \frac{r}{R_0} \right) \quad (2)$$

where $R_0' = R_0(1 + (\beta^2/\alpha^2)R_s/R_x)$ and $\gamma = R_s/(R_s + \alpha^2 R_0')$. In Figure 3c, the electrodes are connected in series with the excitation and sensing circuits. The configuration is effectively the same as the Fourcin's configuration without the guard rings, and the output is given by (1).

Fourcin [11] used a slicing amplifier with a synchronous detector for demodulation of the voltage output from the impedance sensor module. It results in good sensitivity and rejection of external noise. However, the slicing amplifier increases the carrier ripple in the output, requiring a more severe lowpass filtering of the output. Rothenberg [12] used a diode-based peak detector. It reduces the carrier ripple but is more prone to external noise.

After examining these techniques for impedance glottography, we have developed an impedance detector with a wide-band response for sensing of the impedance variation.

3. CIRCUIT DESCRIPTION

Current source excitation is generally preferred in measurements involving biological tissues. Another advantage of this method is that the output voltage is directly proportional to the variation in the glottal impedance. Use of electrodes with guard rings around the central disc electrodes requires a dry skin surface in order to avoid surface leakage of the current between the central disc and its guard ring. But the use of dry electrodes often results in poor contact with the skin around the thyroid cartilage. Further, most circuits use two transformers, one in the excitation circuit and one in the sensing circuit [4,11,12]. Our experiments showed that impedance sensing can be carried out using current excitation and a pair of disc electrodes without guard rings, connected to the secondary of a single high-frequency transformer. Disc electrodes without guard rings avoid the decrease in sensitivity caused by surface leakage. We have used 24 mm diameter stainless steel electrodes, held in place with an adjustable band. If needed, an electrolyte gel may be applied to the skin below the electrodes, to improve the skin-electrode contact. It is also possible to use flexible conducting polymer electrodes with electrolyte gel and adhesives. For amplitude demodulation, we have used a full-wave rectifier–based detector, which is better than the peak detector in noise rejection and the slicing amplifier–based detector in carrier rejection.

A schematic representation of our impedance sensor module is given in Figure 3d, with the output given as

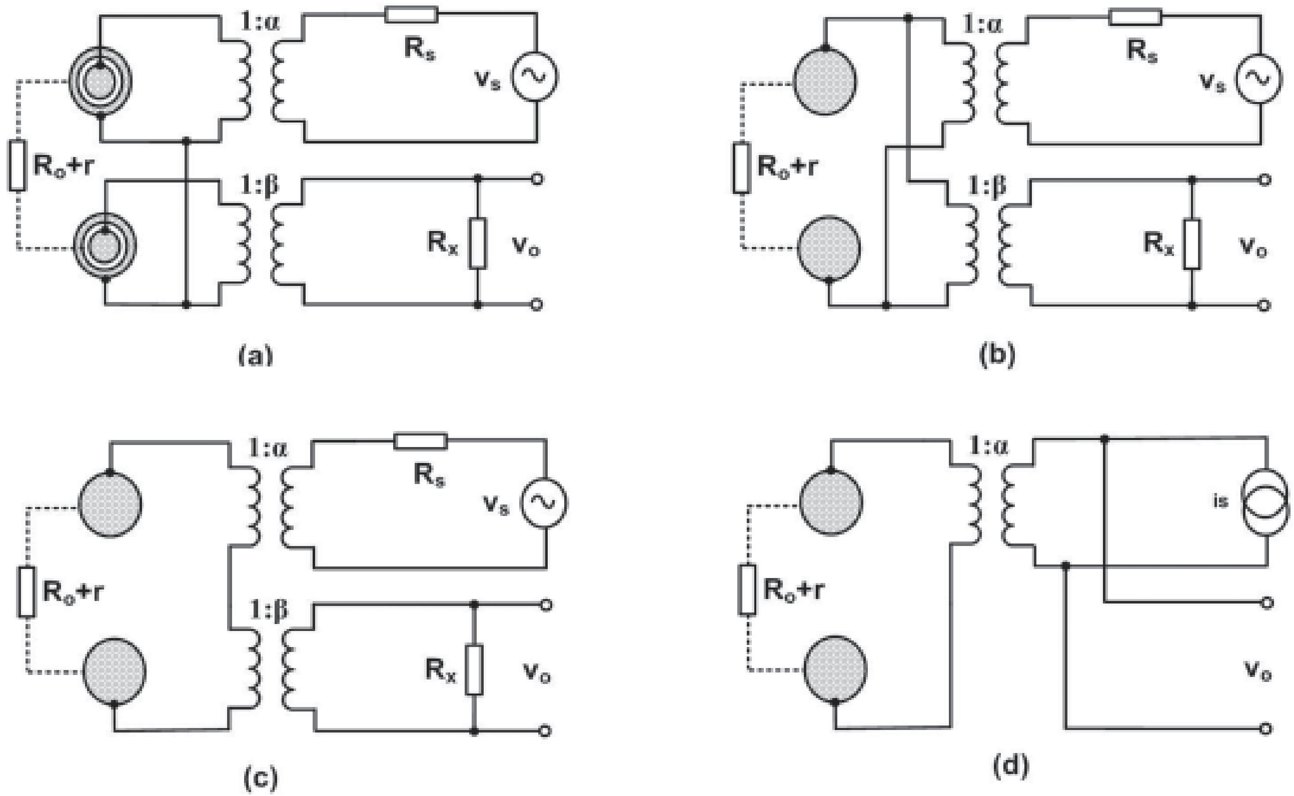

Figure 3: Schematic representation of different impedance detectors: (a) Fourcin's impedance detector [11], (b) Rothenberg's impedance detector using parallel connection of electrodes [12], (c) Rothenberg's impedance detector using series connection of electrodes [12] and (d) current excitation–based impedance detector.

$$v_o = \alpha^2 i_s R_0 \left(1 + r/R_0\right) \quad (3)$$

Use of the current excitation has resulted in higher sensitivity, as the variable component in the output voltage depends only on r/R_0. The impedance sensor module consists of sinusoidal oscillator, voltage-to-current converter and voltage amplifier. The output of the current source is applied to the electrodes held in contact with the skin on both sides of the thyroid cartilage. Impedance variation across the electrodes results in an amplitude-modulated voltage waveform, which is amplified and given to the AM detector module. We have opted for an excitation frequency of 400 kHz and current with rms value of about 1 mA. The circuits of the two modules are described in the following subsections. Both modules work with ±5 V supplies.

3.1 Impedance Sensor Module

Impedance sensor module, as shown in Figure 4, consists of a sinusoidal source and a voltage-to-current converter. Short-time amplitude instability in the source results in noise in the demodulated output. After investigating several sinusoidal oscillator circuits [13,14], we have used function generator IC MAX038–based circuit as the sinusoidal source (measured short-term instability <1.6%). The 2 V peak-to-peak buffered output V_b, from U2, is given to voltage-to-current converter, realized using op amp U3. The current output is given as $I_e = V_b/R_2$, with R_2 = 560 Ω, I_e = 1.28 mA rms. The pair of electrodes E1 and E2 is connected to the current source through high-frequency transformer T1 to provide galvanic isolation. The balanced connection through the transformer also helps in reducing the stray current to ground and the external pickup. Resistance R_3 restricts the zero frequency gain of the circuit. Due to variation in the glottal impedance, the voltage V_c gets amplitude modulated. It is highpass filtered and amplified using op amp U4 to give output V_d.

3.2 AM Detector Module

AM detector circuits using precision rectifier circuits, synchronous detector and slicing amplifier with multiplier have been reported [7,8]. An investigation of diode-based full-wave precision rectifier circuits [14,15] and AD844 current conveyor–based rectifier circuit [16] showed that dynamic range and linearity at the excitation frequency were restricted by the slew rate of op amps. An absolute-value circuit

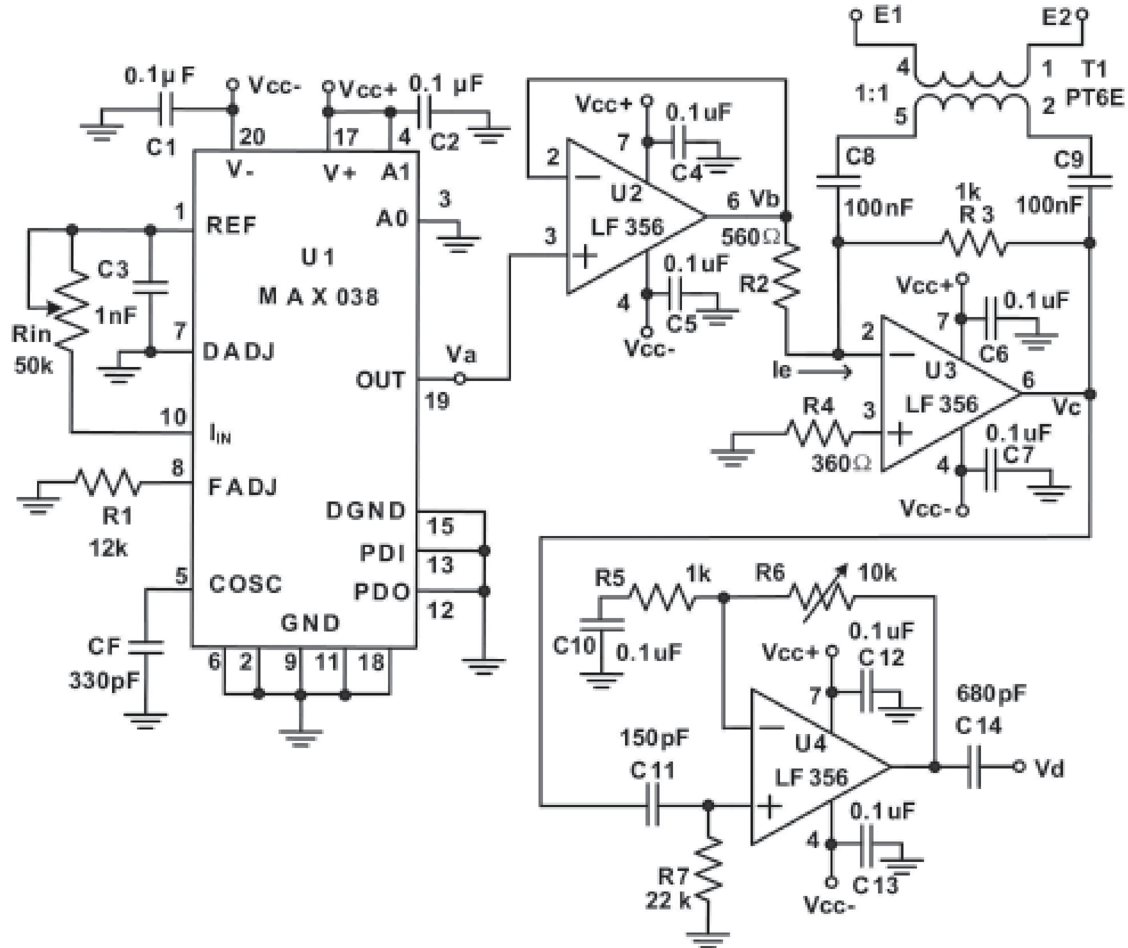

Figure 4: Schematic of the impedance sensor module.

based on voltage feedback clamping amplifier AD8037 was found to provide satisfactory operation. The AM detector module is shown in Figure 5. Clamping amplifier U5, with input V_d connected to lower clamping pin V_L and higher clamping pin V_H connected to V_{CC+}, produces absolute value of the input. The filter after the absolute-value circuit rejects the offset related to basal impedance and high-frequency noise. The EGG waveform typically extends over 75 Hz to 9 kHz. Because of difficulty in designing filters with sharp roll-off and approximately linear-phase shift, it was decided to set the lower and upper cut-off frequencies as 5 Hz and 9 kHz, respectively. The filter was realized by cascading a second-order active Butterworth highpass filter with cut-off frequency of 5 Hz using op amp U6 and fourth-order Butterworth lowpass filter using active filter U7 (MAX074) with a cut-off frequency of 9 kHz. The output was further amplified by U8. The filter showed a nearly flat magnitude response in the 20 Hz to 7 kHz band, and nearly linear-phase response in 60 Hz-1 kHz band.

4. RESULTS

The EGG waveform and speech signal (from a microphone and amplifier) were simultaneously acquired using 2-channel line input of the PC sound card. The instrument was used for EGG recordings from several subjects with normal health and no known laryngeal diseases. It was verified that the waveform for a speaker remained approximately the same for different vowels. An example of the recording is shown in Figure 6, for sustained vowel /*a*/ for a male speaker. The instrument has been also used for long-duration recordings for analyzing voice pitch and studying multiple-period histograms. The results have excellent match with those obtained using another instrument (F-J Electronics EG-90).

The sensitivity and the dynamic response of the glottal impedance detector were measured using a glottal impedance simulator [17] with variable base resistance and a small switching resistance. The glottal impedance detector was found to work satisfactorily for base resistance of up to 170 Ω with a sensitivity of 1 V/ Ω for the time-varying

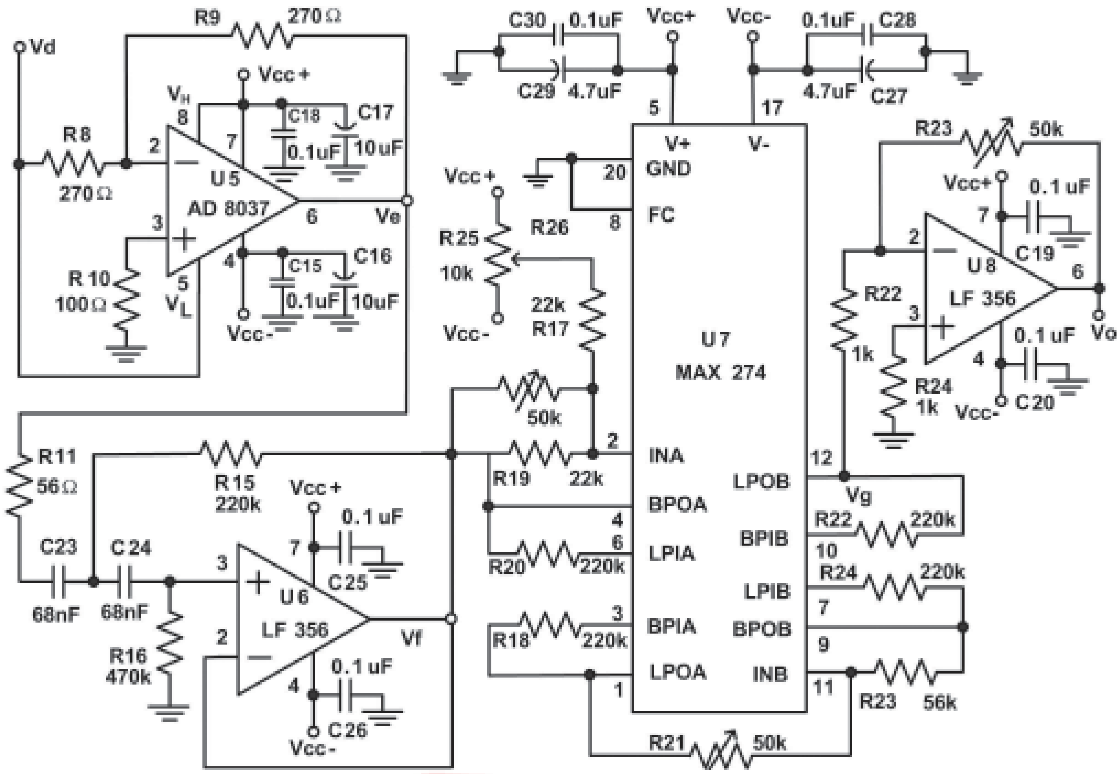

Figure 5: Schematic of the AM detector module.

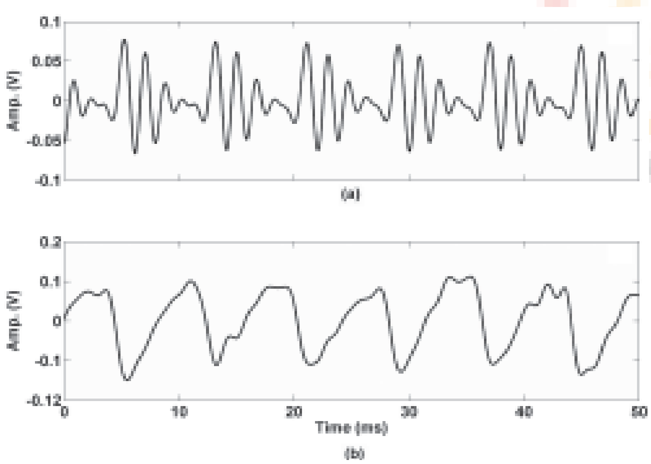

Figure 6: Speech signal and EGG waveform for sustained vowel /a/ for a male speaker: (a) speech signal and (b) EGG waveform.

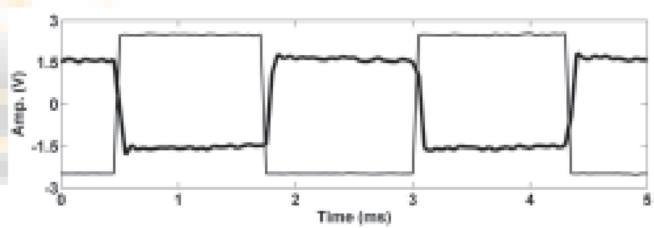

Figure 7: Test waveforms obtained using the impedance glottograph with a glottal impedance simulator, for frequency *Fo* = 400 Hz. The lighter trace indicates the impedance change control signal in the simulator, and the darker trace is the output waveform.

component of the impedance. An example of recorded output is shown in Figure 7. For square wave variation of the resistance in the frequency range of 50 to 400 Hz, the output did not show any appreciable distortion, indicating the magnitude and phase response to be adequate for a faithful sensing of the glottal impedance signal.

5. CONCLUSIONS

An impedance detector for glottography has been developed after an examination of various issues involved. In order to set an upper limit on the current through the biological tissue, it uses current source–based impedance sensing. Disc electrodes without guard rings have been used in order to reduce the problems associated with surface leakage and placement of electrodes. Amplitude detector circuit has been developed using voltage feedback clamping amplifier and filters in order to improve the response and reduce the number of components. Its use for analyzing the voice pitch has been verified with respect to a commercially available instrument, and its sensitivity and the dynamic response have been tested using a glottal impedance simulator. A high-impedance detector circuit may be included in the instrument to indicate improper skin-electrode contact.

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