Parameter Estimation of Respiratory Impedance Measured by Forced Complex-wave Oscillations

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Respiratory impedance data measured using the forced oscillation method were analyzed using an equivalent circuit model and the following conclusions were obtained:
1) Use of the complex wave oscillation method provided a measurement range of over 9 cmH₂O/1/s with a resolution of 0.13 cmH₂O/1/s. 2) The respiratory system was expressed as a two-compartment model. By expressing the circuit characteristics as phasor loci, a method was shown whereby individual impedance parameter values could be derived from the measured impedance phasor loci. 3) Under enflurane anesthesia, the model parameter values were derived as R₁ = 1.2, R₂ = 1.6 cmH₂O/1/s, C₁ = 0.02, C₂ = 0.041/cmH₂O and L = 0.065 cmH₂O/1/s².

(Key Words: forced oscillation, respiratory impedance, respiratory models, frequency dependence of impedance, impedance phasor)

INTRODUCTION

Studies to clarify respiratory mechanics using sinusoidal pressure oscillations imposed through the mouth were reported by Dubois et al in 1956 (4). The impedance is defined as the ratio of the applied pressure to the induced flow, and by varying the frequency, the characteristics of the impedance were determined. The resistive component of the impedance (real part) was found to be in agreement with the respiratory resistance values determined by the body plethysmography and balloon methods. There are many reports related the degree of severity of respiratory diseases as a function of the frequency of applied oscillation, and this is the basis of the forced oscillation method (7,8,12,18). The frequency characteristics of impedance have also been studied through simulation using a mechanical model and its equivalent electrical circuit. These studies have further clarified much physiological data.

This article introduces a modified forced oscillation method using complex wave oscillations. A two-compartment model was employed to express the frequency characteristics of the measured respiratory system impedance values as the loci of the impedance phasors for the model. The method of estimating the values of the model elements by analyses of the measured impedance phasors is also discussed. Based on these observations, impedance analyses of patients under enflurane anesthesia are reported.

MATERIALS AND METHODS

1. Forced Oscillation Method

A block diagram of the prototype measurement system is shown in Fig. 1. A 58 cm diameter Coral loudspeaker serving as the pressure wave source was driven by a complex wave generator. The generator consists of Wienbridge oscillators operating at 1, 2, 3..., 20 Hz and an adder. Oscillation pressure was measured by a pressure transducer (Validyne MP45); induced flow was measured by an identical transducer mounted on a Meliam resistor tube (45 mm in diameter, 0.5 cmH₂O/1/s). Fig. 2 shows an example of oscillation pressure and induced flow waveforms. These signals were passed through a low-pass filter (25 Hz bandwidth and...
24dB/oct. slope) followed by a 12-bit A/D converter, and read in by a data processing unit. The data processing unit consisted of a microcomputer (Z80A) with D/A converters, X/Y plotter, printer and pen oscillograph. Software consisted of window functions (cosine taper), Fourier transformation (FFT), and averaging and display programs as packages. Data processing is described in section 3.

Fig. 1 Schematic diagram of the system. The complex-wave generator consists of 12 Wienbridge-type oscillators and the adder. Its output drives the loudspeaker and generates oscillatory air flow. Pressure imposed on patients and induced flow are converted into electric signals by pressure transducers and read in by the data processing unit via the anti-aliasing lowpass filter and the multiplexing A/D converter. For the data processing, refer to the text.

(1) ECG

(2) PRESSURE

(3) FLOW

Fig. 2 Example of simultaneous recordings of imposed pressure, induced flow and ECG.
2. Respiratory Equivalent Circuit

As indicated in Fig. 3, the frequency characteristics of the respiratory impedance for healthy lungs showed a concave curve with a minimum at 4 ~ 6 Hz. Now let us consider an impedance model with a resistor R, inductor L and capacitor C connected in series. If we assume the respiratory system to be a union of viscoelastic bodies, its behavior can be expressed as follows (15):

\[ P = L \frac{d^2V}{dt^2} + R \frac{dV}{dt} + \frac{1}{C}V, \quad (1) \]

where, 
- \( P \): Pressure imposed at the mouth
- \( L \): Inertial resistance
- \( V \): Air flow
- \( R \): Overall viscous resistance
- \( C \): Overall compliance
- \( dt \): Time derivative

This corresponds to the voltage and current relationship of an RLC circuit. If we let the driving point impedance be \( Z_d \), then:

\[ Z_d = R + j\{\omega L - 1/(\omega C)\} \quad (2) \]

is obtained. When \( \omega^2 = 1/(LC) \), \( Z_d \) is a minimum and is equal to the pure resistance \( R \) which is equivalent to the total respiratory resistance. This relationship provides a basis for the respiratory resistance measurements made by the forced oscillation method.

However, there are cases where the impedance frequency characteristics do not show a minimum; i.e., when the real part is not zero, and it does not remain constant as required by equation (2). As the frequency increases, the real part gradually decreases. This phenomenon is particularly evident in diseased lungs. In addition, because there is no clear correspondence between the respiratory anatomy and model elements, investigation of impedance variations using the model is difficult. To investigate these variations, we used the model shown in Fig. 4. Large airways are assumed to be pipes with rigid walls and their impedance is expressed as \( Z_c \) with a resistor \( R_1 \) and an in-
ductor $L$ connected in series. Small airways that contribute to gas exchange are compliant and expressed as a capacitor $C_1$. Alveolar impedance is expressed as $Z_d$ with a resistor $R_2$ and a capacitor $C_2$ connected in series. $Z_d$ is grouped with the impedance of $C_1$ and expressed as $Z_p$. $Z_t$ is the equivalent impedance of the endotracheal tube used during anesthesia (discussed later). Expressing the forced oscillation frequency as $\omega$:

$Z_d = Z_t + Z_r$, \hspace{1cm} (3)

$Z_r = Z_c + Z_p$, \hspace{1cm} (4)

$Z_c = R_1 + j\omega L$, \hspace{1cm} (5)

$Z_p = \frac{\omega R_2 C_2^2 - j(\omega^2 R_2^2 C_1 C_2^2 + (C_1 + C_2))}{\omega(\omega^2 R_2^2 C_1^2 C_2^2 + (C_1 + C_2)^2)}$ \hspace{1cm} (6)

Each impedance element in the model can be assigned a phasor in the complex plane through which the solution may be constructed graphically to give a clear picture of the entire model. Fig. 5 illustrates this process. (See appendix for a description of the analytical approach.) It also shows the loci of $Z_r$, $Z_c$, and $Z_p$ as functions of $\omega$ in the complex plane. ($Z_d$ and $Z_t$ are not shown.)

We will now discuss the procedure used in deriving the values of the model elements. Expressing the real part of $Z_p$ as $R_e(Z_p)$:

$R_e(Z_p) = \max \{ R_2 C_1^2(\omega^2 + \omega_0^2) \}^{-1}$

$I_m(Z_p) = \frac{\omega^2}{C_1 + \omega_0^2} - \frac{1}{(C_1 + C_2)}$

\hspace{1cm} (7)

where, $\omega_0 = (C_1 + C_2)/(R_2 C_1 C_2)$

and when $\omega = \omega_0$;

$R_e(Z_p) = (1/2)R_2(1 + m)^{-2}$

$I_m(Z_p) = (1/2)R_2(1 + m)^{-2}(1 + 2m)$

$\omega_0 = (1 + m)/(R_2 C_1)$

\hspace{1cm} (10)

where, $m = C_1/C_2$

and when $\omega = 0$;

$R_e(Z_p) = R_2(1 + m)^{-2}$.

From the above equations we can see that $R_e(Z_p)$ at $\omega = \omega_0$ is half that at $\omega = 0$, and provided that this holds true at the lower limit.
Fig. 5  Frequency characteristics of the circuit shown in Fig. 4 represented as phasor loci in the complex plane. The vertical and horizontal axes shown as JX-JX and R indicate the imaginary and resistive components of impedances. The frequency characteristics of \( Z_p \), \( Z_c \) and \( Z_r \) are obtained as the phasor loci shown as \( Z_p(\omega) \), \( Z_c(\omega) \) and \( Z_r(\omega) \), respectively. With an increase in frequency, \( Z_p \) and \( Z_c \) approach zero and infinity, respectively, and \( Z_r \) gradually approaches \( Z_c \). The intersection of the locus of \( Z_r \) and the real axis corresponds to the resonant point or the resonant frequency. Note that this point does not correspond to the point of the least magnitude expressed as the distance from the origin \( O \) to the loci. See text for details.
of our system \((\omega = 1 \text{Hz})\), then equations (9)\textendash(11) can be solved for \(C_1\), \(C_2\) and \(R_2\).

3. Data Processing

Considering the driving point impedance \(Z_d(\omega)\) at the mouth, we let the power spectrum of the applied pressure and induced flow be \(S_p(\omega)\) and \(S_t(\omega)\), and the cross spectrum and phase be \(X(\omega)\) and \(\phi(\omega)\), respectively, to give the following relation in which the phase difference between pressure and flow is equal to \(\phi(\omega)\) (1,10):

\[
Z_d(\omega) = \frac{|S_p(\omega)|}{|S_t(\omega)|} = \frac{S_p(\omega)}{|X(\omega)|}.
\]

This spectral calculation is performed by an FFT algorithm for which the following parameters are required:

\[
\Delta t = T/N, \ \Delta f = 1/T, \ \text{fmax} = 1/2(\Delta t) = N/2T,
\]

where, \(\Delta t\): Sampling interval,

\(N\): Number of samples taken,

\(\Delta f\): Frequency unit,

\(T\): Sampling time,

\(\text{fmax}\): Maximum frequency.

In our system \(\text{fmax} = 20 \text{Hz}\) and \(\Delta f = 1 \text{Hz}\) are applicable but to take into account the influence of secondary oscillation sources in the respiratory system (such as heart beat and spontaneous breathing), we chose \(\Delta f = 0.2 \text{Hz}\), \(\text{fmax} = 25 \text{Hz}\), \(\Delta t = 20 \text{msec}\) and \(N = 256\). Spectral estimation error was minimized by statistical averaging and ensemble averaging to permit a data read-in time of about 15 seconds. A low-pass filter was used to control the bandwidth and provide spectral compensation through use of the cosine taper window.

4. Method

Four Meliam resistance tubes (length: 2 \(\sim\) 5 cm, 0.25 \(\sim\) 2.6 cmH\(_2\)O/1/s) were prepared and calibrated with the calibration analyzer (RT-200, Tymeter), and then connected to the system to evaluate system resolution and dynamic range, as well as to calibrate impedance during measurement. Two types of endotracheal tubes were then connected to the system (ID 8 \(\sim\) 9 mm, length 32 cm, Portex), and the results used to develop an equivalent circuit for the tubes. Equation (3) was used to compensate for impedance changes caused by the endotracheal tube.

Six patients aged 26 \(\sim\) 42 years and weighing 40 \(\sim\) 65 kg with no respiratory or circulatory abnormalities were tested under N\(_2\)O-O\(_2\) and enflurane anesthesia. Measurements were made with the patients lying face-up and were repeated three times over about 15 seconds in the anemic condition. Measured pressure and flow were recorded alongside ECG and arterial pressure on a data recorder (A-67, Sony). Analysis was then performed using the data processing system described in section 3. In some cases, a resistance tube (1.5 cmH\(_2\)O/1/s) was inserted between the endotracheal tube and the system.

RESULTS

Table 1 shows the frequency characteristics of the resistance tubes relative to the 1 Hz value of the 2.6 cmH\(_2\)O/1/s tube. The higher resistance values were obtained by combining two or more of the tubes. The lower figures in the frequency columns are the combined resistances with a nominally 0.25 cmH\(_2\)O/1/s tube in series. The measurement error at 1 Hz was within \(\pm 3.6\%\). The increases in resistance were at least 56\% of the nominal added resistance. This indicated that within a dynamic range of 9 cmH\(_2\)O/1/s, the system had a maximum resolution of 0.13 cmH\(_2\)O/1/s. When more than two resistance tubes were combined, the capacity in the direction of the tube length increased, resulting in a markedly larger frequency dependence for the higher resistance values.

Fig. 6 shows the frequency characteristics of the endotracheal tubes. The phasor locus moved in parallel to the imaginary axis as the frequency increased, showing that the impedance may be expressed as \(Z_t = R_t + j\omega L_t\). From the figure we can see that the two tubes have constants of \(R_t = 1.6\) and 2.2 cmH\(_2\)O/1/s and \(L_t = 0.056\) and 0.072 cmH\(_2\)O/1/s\(^2\), respectively. Above 10 Hz, there was a tendency for an increase in resistance and the impedance no longer followed the above one dimensional model. This increase appeared to result from turbulence generated over that frequency band as will be mentioned in the Discussion, and it now suffices to say that compensation coefficients were applied to the above constants.
Table 1  Frequency characteristics of fixed resistor tubes to demonstrate the resolution power of the system. All values are calibrated relative to the 1Hz value of the 2.6 cmH₂O/1/s resistor tube. The lower values in the frequency columns are the combined resistances with a nominally 0.25 cmH₂O/1/s resistor tube in series. See text for further details.

<table>
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<th>Resistance length</th>
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<th>3.5</th>
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Resistance: cmH₂O/1/s
Length: mm
Frequency: Hz

Fig. 6  Frequency characteristics of the endotracheal tubes represented as phasor loci in the complex plane. The phasor locus moves in parallel to the imaginary axis (shown as the reactance component) as the frequency increases, showing that the tube impedance may be expressed as \( Z_t = R_t + jωL_t \). The intersection of the locus and the real axis (shown as the resistance component) corresponds to the resistance measured by constant flow and pressure difference.
The frequency characteristics of the real and imaginary parts of the respiratory impedance under anesthesia are shown in Fig. 7. Both real and imaginary parts were found to have an irregular region at 6 ~ 8 Hz, but in general, the impedance closely followed the phasor loci of $Z_r(\omega)$ in Fig. 5. The minimum magnitude appeared an average value of 12.6 Hz. The reproducibility of the data was within ±0.6 cm H$_2$O/1/s. The change in impedance when a resistance tube (1.5 cm H$_2$O/1/s) was inserted is shown in Fig. 8. The locus of $Z_r(\omega)$ underwent a shift in the real axis direction at all frequencies without any change in $Z_p(\omega)$. Therefore, the insertion of an external resistance tube corresponded to an increase in $R_1$ of $Z_c(\omega)$.

Table 2 shows the values for the model elements derived as described in section 2. When a resistance tube was inserted (+ 1.5 standard), the resulting increase in $R_1$ was measured at 1.2 cm H$_2$O/1/s, in close agreement with Fig. 8.

![Fig. 7](image-url) Frequency characteristics of the resistance and reactance components of the respiratory impedance for each patient under anesthesia. The vertical scale runs from -8 to 5 cm H$_2$O/1/s and the horizontal scale is in Hz. Both traces have an irregular region at 6 ~ 8 Hz due to cardiogenic oscillations, but in general, closely follow the phasor loci of $Z_r(\omega)$ in Fig. 5. Note that with an increase in frequency, the resistance components gradually decrease, while the reactance components go upward from capacitive (negative) to inductive (positive).
Fig. 8 Frequency characteristics of the respiratory impedance with added external resistance as phasor loci in the complex plane. +STD1.5 indicates the loci with an added external resistor of 1.5 cmH$_2$O/1/s. The loci of $Z_p$ were derived as described in section 2. The locus of $Z_r$ undergoes a parallel shift in the real axis without any change in $Z_p$. This shift corresponds to an increase in $R_1$ of $Z_c$ in Figs. 4 and 5.

Table 2 Estimated values of the model in Fig. 4 for patients under anesthesia. The bottom column shown as #1 + STD1.5 indicates the case with an external resistor of 1.5 cmH$_2$O/1/s added to case #1. For the symbols, see the text.

<table>
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<tr>
<th>PATIENT</th>
<th>ID.</th>
<th>R1</th>
<th>L</th>
<th>R2</th>
<th>C1</th>
<th>C2</th>
<th>$\omega_0$</th>
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<td>5040N</td>
<td>0.30</td>
<td>0.080</td>
<td>1.59</td>
<td>0.013</td>
<td>0.040</td>
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</tr>
<tr>
<td>#2</td>
<td>417HT</td>
<td>1.11</td>
<td>0.078</td>
<td>1.58</td>
<td>0.014</td>
<td>0.049</td>
<td>9.0</td>
</tr>
<tr>
<td>#3</td>
<td>3251M</td>
<td>0.92</td>
<td>0.070</td>
<td>0.99</td>
<td>0.029</td>
<td>0.038</td>
<td>9.0</td>
</tr>
<tr>
<td>#4</td>
<td>428MM</td>
<td>3.40</td>
<td>0.029</td>
<td>2.28</td>
<td>0.016</td>
<td>0.030</td>
<td>7.0</td>
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<td>4270M</td>
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<td>0.055</td>
<td>1.63</td>
<td>0.013</td>
<td>0.031</td>
<td>11.0</td>
</tr>
<tr>
<td>#6</td>
<td>428KT</td>
<td>1.04</td>
<td>0.080</td>
<td>1.47</td>
<td>0.015</td>
<td>0.055</td>
<td>9.0</td>
</tr>
<tr>
<td>MEAN</td>
<td></td>
<td>1.19</td>
<td>0.065</td>
<td>1.59</td>
<td>0.017</td>
<td>0.040</td>
<td>9.2</td>
</tr>
<tr>
<td>#1+</td>
<td>STD1.5</td>
<td>1.53</td>
<td>0.078</td>
<td>1.49</td>
<td>0.014</td>
<td>0.046</td>
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</tr>
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</table>

R1, R2: cm H$_2$O/1/s,  L: cm H$_2$O/1/s$^2$
C1, C2: l/cm H$_2$O,  $\omega_0$: Hz
DISCUSSION

The original method developed by DuBois et al. (4) used a motor driven diaphragm. Grimby et al. (12) first suggested a method based on a loudspeaker which proved successful, resulting in a large amount of clinical data. Further experiments by Michaelson (17) and Landser et al. (18) involved the development of new methods using the speaker to produce random or impulse noise, and spectral analysis to calculate frequency characteristics. These measurements were performed over extremely short intervals and offered the advantage of providing the frequency characteristics as a continuous spectrum. The unit power at a given spectral value was low, however, making the measurements vulnerable to the influence of ambient noise sources and vibrations generated within the body itself. For this reason, in the lower frequency band it is very difficult to evaluate variations in compliance, which are the main components of the impedance. Our method was an expansion of the one suggested by Landser (13): the oscillation source was designed to use the minimum required spectrum number with preemphasis on the lower frequency band. As a result, stable impedance calculation was possible without a reduction in coherence (a value expressing the coupling of applied pressure and induced flow). In our opinion, the complex wave oscillation technique combined with spectral analysis offers an effective tool for the investigation of respiratory functions.

The respiratory model shown in Fig. 4 is an expansion of the small airway model developed by Mead (16) and is identical to the circuit model developed by Blesser (2). This is an advanced model that matches the respiratory anatomy. We were able to explain the circuit model as phasor loci in the complex plane as shown in Fig. 5. Impedance is a two-dimensional quantity that may be described by magnitude and phase in the complex plane. By relating the model's overall phasor $Z_r$ to its component phasors, analysis of parameter variations was greatly simplified. If, for example, we want to see the effect of a change in $R_1$ on $Z_r$, it is expressed as the phasor locus parallel to the imaginary axis shown in Fig. 9. The results in Fig. 8 can then be seen as clearly resulting from an increase in $R_1$. This circuit model has been solved analytically, and numerous articles based on computer simulations were published (3,12,16,18). They shed light on the relationship between parameter variations and diseases, but have not allowed full analysis of continuously varying parameters. For example, variations of separate parameters may result in identical effects. At present, this method is performed by a computer program with the phasor loci displayed on a CRT.

The 8 mm diameter tube impedance shown in Fig. 6 had a resistance of 2.2 cmH2O/1/s, 67% of the 3.3 cmH2O/1/s measured under respiratory flow by Macintosh et al (14). The air flow during respiration is not laminar (as stated in their report): therefore, allowing for an increase due to turbulent resistance, the value we obtained is not excessively low. However, since we used complex wave oscillations up to 20 Hz, we may postulate that turbulence was generated at high frequencies although no report has yet dealt with the effect of turbulent flow on respiratory impedance. Studies on high frequency ventilation have concluded that turbulence is generated at the airway bifurcations (9). Our data were weighted for the turbulent effect in accordance with the airway diameter, but further research in this area is still required.

The minimum impedance value shown in Fig. 7 occurred at 12.6 Hz, a higher value than suggested by other studies (4,7,8,12,18). This resulted from the use of tube impedance compensation and the reduced effect of the upper airway inductance. Tsai et al(23) have reported similar results.

Since DuBois (4), several methods have been devised to derive values for model elements from the frequency characteristics. Pimmel et al (19) and Slutsky et al (20) expressed the ratio of $R_1$ and $R_2$ in Fig. 4 as $F_p$. They discussed the function transformation techniques to derive $F_p$, and have also reported on an algorithm for direct derivation of parameter values using regression analysis (5,6,29). These methods are prone to modeling errors (20) in deriving a model equation, and the algorithm constructed requires a long computing time. There may be cases where a solution can not be obtained (5,6). In our method, we assumed that at the upper and lower frequency limits,
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Fig. 9  Effect of the increase of $R_1$ on the respiratory impedance $Z_T$ (frequency fixed). $Z_r$ and $Z_c$ move to the right uniformly parallel to the real axis with an increase in $R_1$, while $Z_p$ remains constant. The origin of $Z_r$ shown as $R_1=0$ is determined as the diagonal of the parallelogram with fixed sides, $\omega L$ and $Z_p$.

$Z_c(20\text{Hz}) = Z_r(20\text{Hz})$ and $R_c(Z_p)(1\text{Hz}) = R_r(Z_p)(0\text{Hz})$. Using data from Spells (21) and Bobbaers et al (3) to calculate $Z_p(\omega)$, we obtained $Z_c(20\text{Hz}) = 0.9Z_r(20\text{Hz})$ and $R_c(Z_p)(1\text{Hz}) = 0.95R_r(Z_p)(0\text{Hz})$, allowing for roughly 10% error. In view of the limits of the system resolution power and the reproducibility of the data, the method presented here appears to be practical. Although the upper frequency limit can easily be raised, there is no reliable way to measure the air flow; therefore this method will not improve the accuracy.

There have been many publications dealing with small airway compliance $C_1$, peripheral resistance $R_2$ and compliance $C_2$. In subjects with healthy lungs in the sitting position without anesthesia, Mead et al (16) obtained $C_1 = 0.01$, $C_2 = 0.1$ l/cmH$_2$O, $R_2 = 0.5$cmH$_2$O/1/s; and Bessor (2) obtained $C_1 = 0.02$, $C_2 = 0.18$, $R_2 = 0.8$. Measurements performed by Takishima et al (22) in the 8th generation showed $C_1 = 0.02$ and at the periphery $C_2 = 0.04$, $R_2 = 1.59$. The values we obtained for $C_2$ and $R_2$ differ markedly from the above. However, Gold et al (11), who measured patients during surgery under anesthesia, reported that variation in body position (sitting or lying face up) and the depth of anesthesia could result in compliance reductions of 17% and 33%, respectively, in the range 0.02–0.191/cmH$_2$O. These derived values show significant variations due to differences in the mechanical characteristics of the lung periphery as well as differences in measurement conditions. Our results appear to reflect altered lung peripheral characteristics due to the body position and the relaxant and anesthetic used. We hope that continuing studies will enable us to shed light on the relationship between
anesthetic and respiratory functions in future reports.

APPENDIX

Determination of the Respiratory Impedance Phasor

The alveolar impedance phasor \( Z_a \) is obtained from the sum of two phasors, \( R_2 \) and \( 1/(\omega C_2) \). If a straight line perpendicular to \( Z_a \) at the tip intersects the real and imaginary axes at points A and B, respectively, phasor AO and BO are the parallel components of \( Z_a \) on the axes \( Z_a = AO//BO \) where O is the origin. BO and \( 1/(\omega C_1) \) both lie along the imaginary axis and yield the resultant phasor OC constructed by the procedure illustrated in Fig.5 as \( OC = OB//\{1/(\omega C_1)\} \). \( Z_p \) which is a parallel impedance phasor of \( Z_a \) and \( 1/(\omega C_1) \) can be obtained from drawing a line DO perpendicular to the line AC from the origin O (\( Z_p = CO//AO \)). The large airway impedance phasor \( Z_e \) is obtained from the sum of two phasors, \( R_1 \) and \( \omega L \). Finally, the total respiratory impedance phasor \( Z_r \) is determined as a diagonal of the parallelogram with sides \( Z_p \) and \( Z_e \).

REFERENCES