A NEW METHOD FOR MEASURING NASAL CONDUCTIVITY

A. Spoor

Introduction

The present author has given in an other paper (Spoor, 1963) a review of some theoretical considerations about the air flow in the nose. Actual measurements of both pressure and flow have been described by several authors: Cottle, 1963; Drettner, 1961; Malcomson, 1959; Seebohm, 1958; Semerak, 1958; Stoksted, 1953. Different methods are used; measurement of pressure during constant flow and flow measurement during constant pressure. From these measurements the so-called nasal resistance or the inverse the nasal conductivity can be calculated for the conditions set by the experiment.

The aim of the present study was to develop an apparatus that measures both pressure and flow during actual normal breathing and that calculates itself the nasal conductivity during each whole breathing cycle. The result is recorded on paper.

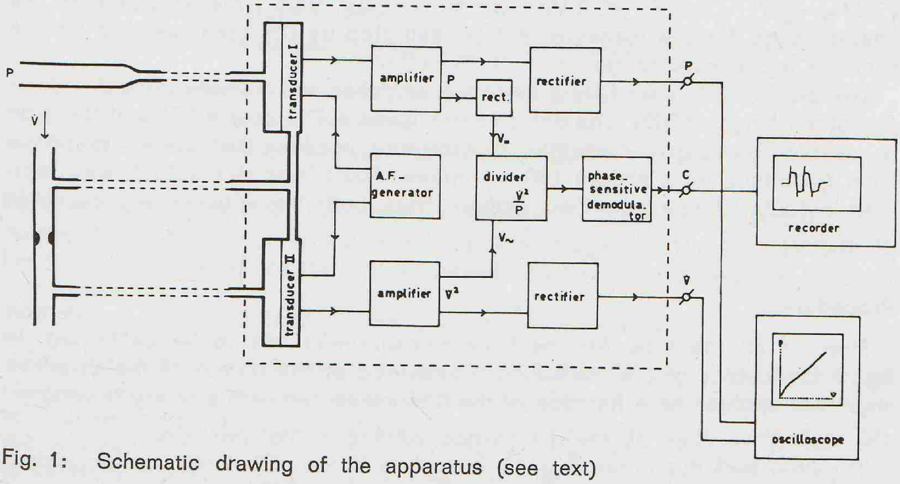

Fig. 1: Schematic drawing of the apparatus (see text)
Dessin schématique de l'appareil (voir le texte).

Apparatus (figure 1)

The apparatus consists of two strain-gauge pressure transducers, Langham and Thomson, type UP 1, differential. Two tubes with close fitting molds are slightly pressed against the nostrils. One tube for the pressure measurement is connected directly to the transducer I and forms a closed system. The other is open at both ends and is used for the flow measurement. In the tube is a narrower part and the pressure developed across this narrowing is measured by transducer II. In order that the system can be used for inspiration and expiration this latter pressure measurement is made symmetrical by the use of two tubings and a differential pressure transducer. The tubing on the side of the nostril is also connected to the other side of transducer I, which is also a differential transducer and that actually measures the pressure difference developed in the nose during breathing.

These transducers are followed by quite a bit of electronic circuitry which is wholly transistorized. First of all the transducers are not fed by direct current but by an alternating current of 1200 cps, produced by an oscillator (Emms, 1960). This is necessary because the following dividing circuit needs alternating current. The disadvantage is that two kinds of balancing of the transducers are necessary: resistance and capacitive balancing. The response of the transducers is now an alternating current, the amplitude of which is proportional to the pressure and the phase is shifted 180° if the sign of the pressure is reversed (inspiration and expiration). The signal from the transducers is amplified selectively in order to suppress noise and hum. The signal from the pressure transducer I is then rectified ($V_{=}$) before going to the divider circuit (Smith, 1960). The other signal $V_{\sim}$ from the "flow" transducer II (also amplified selectively) goes straight to the divider.

The output of the divider is proportional to the quotient $\frac{V_{\sim}}{V_{=}}$ if three conditions are fulfilled.
1 : $V_{\sim} \leq V_{=}$; 2 : $V_{=} < 6$ Volt; 3 : $V_{=} > 0.2$ Volt. This means that the usable range for the measurement is restricted by the pressure, but for the practice a pressure range of 1 : 30 is sufficient.

The divider is followed by a selective amplifier and a phase sensitive demodulator (Wright, 1962). The demodulator gives a DC output (C) and the sign is reversed going from inspiration to expiration because then the AC response from transducer II is shifted 180° in phase. Apart from this output, also rectified outputs of the amplified signals from both transducers are available (P and $\dot{V}$).

Procedure

First of all the tube for the flow measurement has to be calibrated. In fig. 2 the output of the transducer (measured at the output of the selective amplifier) is given as a function of the flow measured with a rotary flowmeter. The flow $\dot{V}$ is given as the discharged volume in liter per min.

Together with this curve is given a curve with the calculated data according to the formula $P = 0.0139\ \dot{V}^2$. There is a very close agreement between both curves and it is reasonable to assume for the following calculation a square law relationship between the output of this transducer and the air-flow through the tube. In other words if $V_{\sim}$ is the output of transducer II (plus amplifier) then $V_{\sim}$ is proportional to $\dot{V}^2$ or $V_{\sim} = C_1\dot{V}^2$ (C_1 constant).

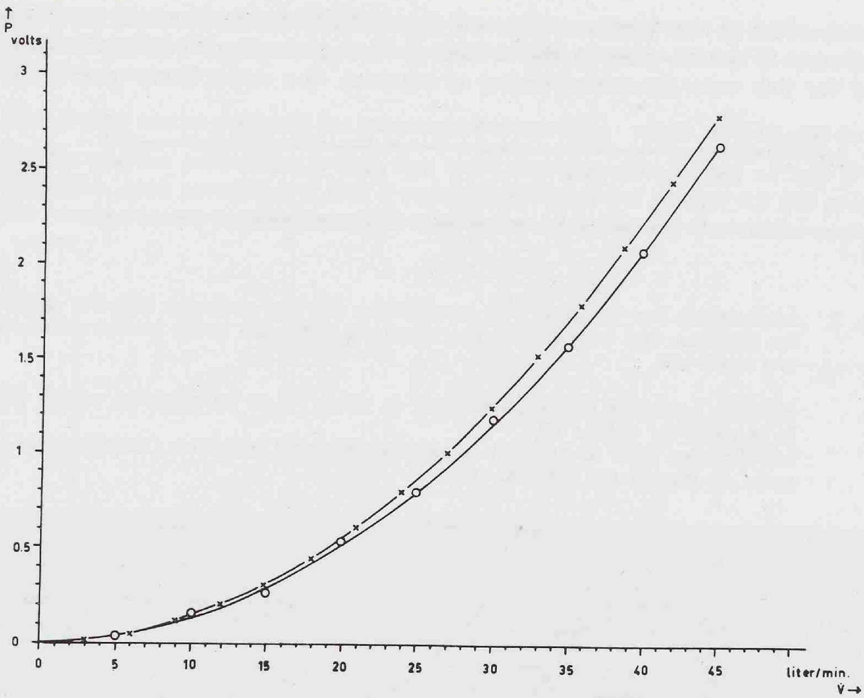

Fig. 2: Calibration of the flowmeter with a rotary flowmeter.
O———O : measured data.
X———X : calculated data according to the formula $P = 0.0139\ \dot{V}^2$.

Etalonnement du débit mètre par un mètre rotatoire.
O———O : valeurs mesurées.
X———X : valeur calculées d'après la formule $P = 0.0139\ \dot{V}^2$.

For the measurement of the pressure across the nose it is valid that the output of transducer I ($V_{=}$) is proportional to that pressure P or $V_{=} = C_2P$. Let we call the output of the divider amplified and demodulated, C^2, then we find that C_2 is proportional to the quotient $\frac{V_{\sim}}{V_{=}}$ or $C^2 = c\frac{\dot{V}}{P}$.

Now we must give a definition for C. If $c = 1$, then C^2 gives the (discharged volume in ccm per second)^2 at a pressure of 1 cm of water across the nose or C gives the discharged volume per second at a pressure of 1 cm of water. We will call C the nasal conductivity. The only thing we have to do now is to calibrate our apparatus in such a way that c is 1 and to make our meter scale according to a square law.

In our apparatus the pressure range goes from 0—4 cm of water and the flow range from 0—1000 ccm per second. The corresponding conductivity scale goes from 0—500 ccm/sec., (fig. 3). The pressure, flow and conductivity can be measured independently. For instance if the pressure is 2 cm of water and the flow is 705 ccm/sec. then the conductivity is 500 ccm/sec. This corresponds to the division by the divider in the apparatus of equal quantities, and for this case the meterreading is maximal. The same holds true for all

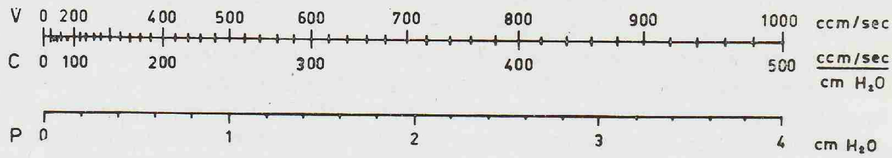

Fig. 3. Meterscales for the flow (V̇), conductivity (C) and pressure (P). The scales for the flow and conductivity are quadratic ones, the scale for the pressure is a linear one.

Echelles graduées de l'appareil pour le débit (V̇) la perméabilité nasale (C) et la perte de charge (P).
Les échelles pour le débit et la perméabilité sont au carré, l'échelle pour la perte de charge est linéaire.

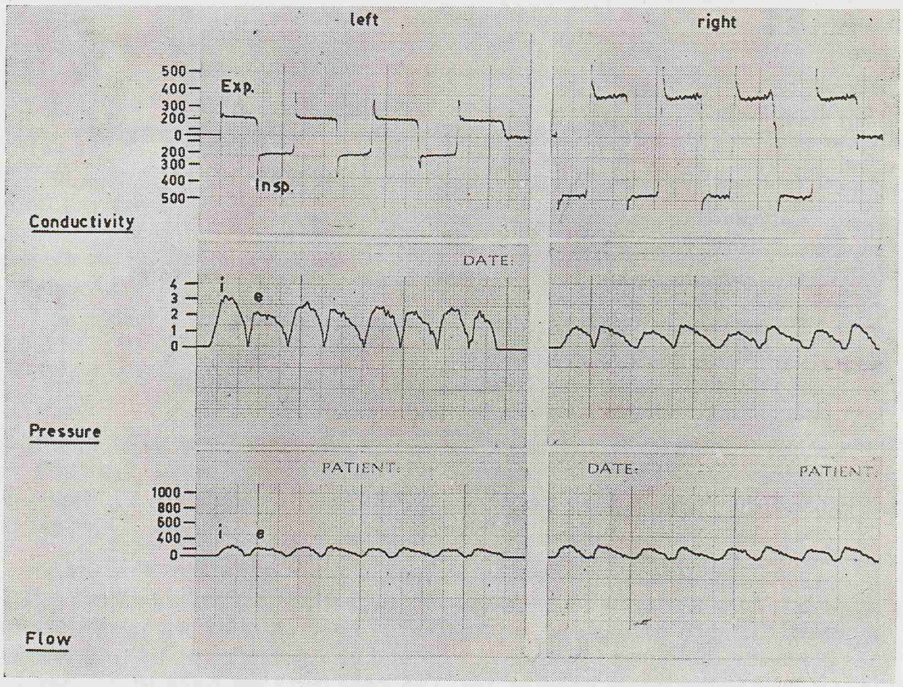

Fig. 4. Recordings of conductivity, pressure and flow in a same run for the left and the right half of the nose. Note the practical constant value of the conductivity during inspiration and expiration.

Enregistrements de la perméabilité nasale, de la perte de charge et du débit dans une même mesure pour les moitiés gauche et droite du nez. Notez la valeur pratiquement constante de la perméabilité pendant l'inspiration et l'expiration.

situations where the meterreadings for P and $\dot{V}$ are equal and in these cases the reading for C is maximal in accordance with what is said in the section on the apparatus.

The divider works correctly for pressures between $\frac{1}{30}$ x4 cm = 0,13 and 4 cm of water and the calibration is such that in all cases condition 1 : $V_{\sim} \angle V_{=}$ is fulfilled. Also conditions 2 and 3 are fulfilled when the pointer for the pressure measurement is between 0,13 and 4 cm of water on the meter scale.

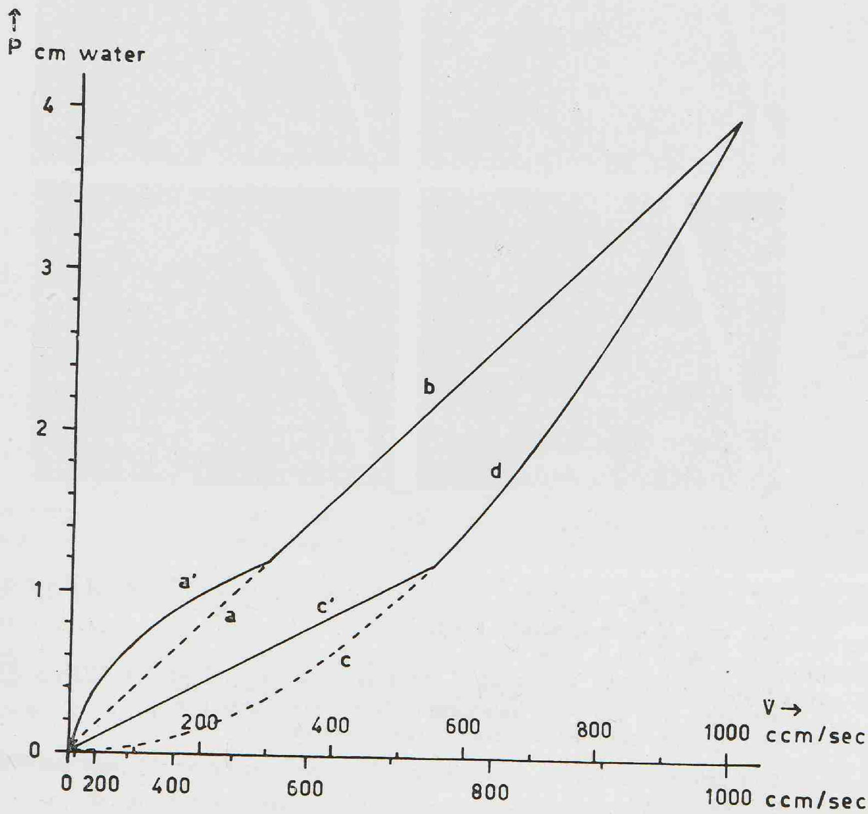

Fig. 5. The function P= c $\dot{V}^2$ for a linear scale of $\dot{V}$ (c and d) and for a quadratic scale of $\dot{V}$ (a and b). Assuming that for low values of $\dot{V}$ ($\dot{V}$ < 550 ccm per second in the drawing) P is a linear function of $\dot{V}$ then we get c' instead of c for the linear scale of $\dot{V}$ and a' instead of a for the quadratic scale of $\dot{V}$.

La fonction P= c$\dot{V}^2$ pour une échelle linéaire de $\dot{V}$ (c et d) et pour une échelle au carré de $\dot{V}$ (a et b). Si nous supposons que pour des valeurs inférieures de $\dot{V}$ ($\dot{V}$ < 550 cm^3 par seconde dans le graphique) P est une fonction linéaire de $\dot{V}$, nous obtenons alors c' au lieu de c pour une échelle linéaire de $\dot{V}$, et a' au lieu de a pour une échelle au carré de $\dot{V}$.

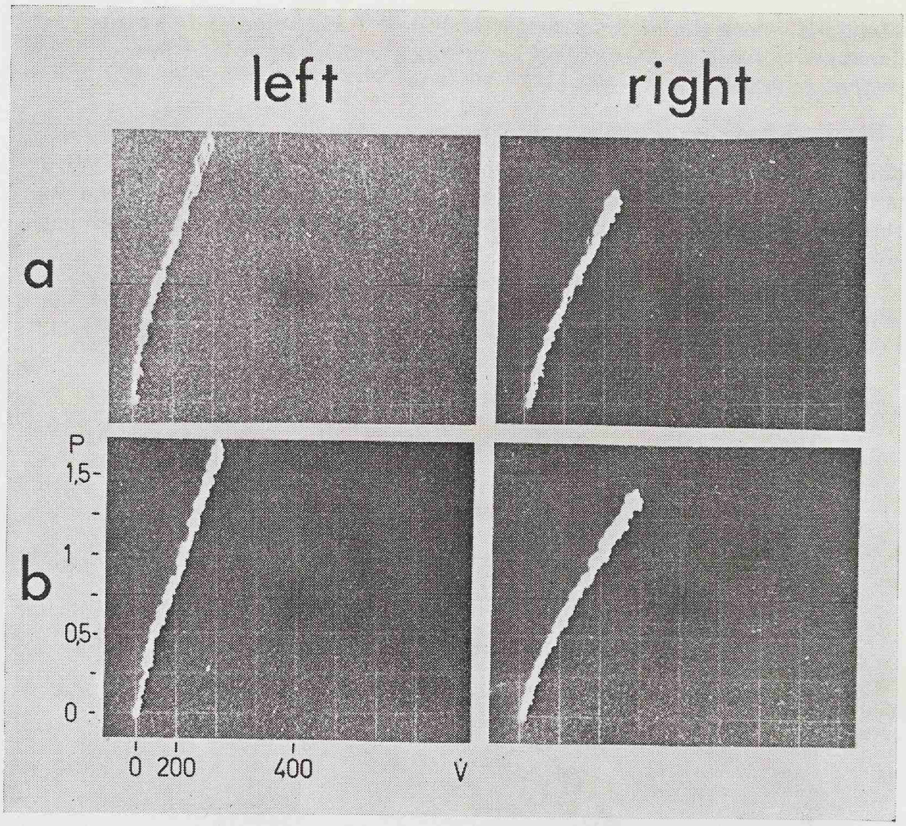

Fig. 6. X-Y recordings of inspiration (a) and expiration (b) for both halves of the nose.
Hor. axis: $\dot{V}$ on a quadratic scale.
Vert. axis: P on a linear scale.
Note that for low values of $\dot{V}$ there is a deviation from the straight line indicating that the P—$\dot{V}$ relationship is a more linear one.
Photographs taken from a storage oscilloscope.

Enregistrements X-Y de l'inspiration (a), et de l'expiration (b) pour les deux moitiés du nez.
Axe horizontal: $\dot{V}$ sur une échelle au carré.
Axe vertical: P sur une échelle linéaire.
Notez qu'il y a pour des valeurs inférieures de $\dot{V}$ une déviation de la ligne droite, qui montre que la relation entre P et $\dot{V}$ est plus linéaire.
Les photos sont faites d'un oscilloscope «emmagasinage».

Results

In figure 4 is given a recording of the "nasal conductivity" measured with the apparatus. During the greater part of the breathing cycle this is a constant figure, thus independent of the actual pressure and flow. The normal course of these quantities is shown in the same figure. Because our apparatus gives the conductivity using the assumption that there is a square law relationship between pressure and flow then reasoning the other way round if the result is a constant figure then this relationship must hold for the air flow in the nose during almost the whole breathing cycle.

Because our apparatus has the possibility to give electrical outputs corresponding with the pressure and the square value of the flow it is possible to lead these outputs to an oscilloscope one to the vertical deflecting plates the other to the horizontal deflecting plates. This has already been suggested by Tonndorf (1958). We must remark that in this case there is no lower limit (except for an unadequately balancing) for the pressure to be recorded. In the conductivity measurement there was such a lower limit of 0.13 cm of water, otherwise the divider would not work properly.

If we put now a linear scale for the pressure (cm of water) to the vertical axis and a quadratic scale for the $\dot{V}$ values to the horizontal axis then in the case of a square law relationship between pressure and flow the produced curve for inspiration and expiration must be a straight line (fig. 5a and b). If the scale to the horizontal axis was linear for $\dot{V}$ then the curve would have the exponential form (fig. 5 c and d). Let we now assume that the pressure-flow relationship is partly linear (for low values of $\dot{V}$) and partly quadratic then the curve (for a linear scale for $\dot{V}$) would look like fig. 5 c' and d. If we transform this curve to the appropriate form for a quadratic scale of $\dot{V}$ its appearance is like a' and b in fig. 5. That means in the beginning it is like an exponential curve with the exponent equal to ½ and from a certain point onwards it is a straight line.

What we actually find is to be seen in fig. 6. These are pressure-flow curves photographed from the oscilloscopescreen during normal breathing. The experiments are performed as follows. The subject is asked to breath quietly and the intensity of a storage oscilloscope is set in such a way that the trace is just stored during inspiration or expiration. In fig. 6 photographs are given of X-Y curves for inspiration and expiration measured at the left and the right nostril. It is clear that there is a very close relationship with the theoretical curve for a partly linear and partly quadratic pressure-flow relationship as given in fig. 5 a' and b.

Conclusion

The results of the measurements described in this paper give quite good evidence for a partly linear, partly quadratic pressure-flow relationship for the air stream through the nose. This is in accordance with the theoretical considerations given in an earlier paper. In that paper the laws governing the air stream through tubes were given and the conclusion was that for a certain diameter of the tube the air stream would be laminar for low values of the flow and turbulent for higher values of the flow.

In the first case a linear relationship is valid for the pressure flow-relationship and in the second case an almost quadratic one. In between there will be a transition range.

In any case for higher values of the flow a transition to a quadratic relationship is very probable because of the influence of in- and out-stream phenomena.

It will be quite clear that if the dimensions of the tube, or the system as in our case the nose, are variable the transition range shifts to lower (system narrower) or higher flow values (system wider) with accordingly lower or higher conductivity.

From the experiments with the "conductivity meter" it follows that for the greater part of the respiratory cycle the flow is above the lower limit for turbulence. Therefore it is useful to define the nasal conductivity as the discharged volume in ccm per second for a pressure of 1 cm of water. This quantity can easily be measured with the described apparatus. The defined nasal conductivity gives to the nose-surgeon a figure which has some advantages. Firstly, conductivities measured for both halves of the nose may be summed directly and this sum gives the behaviour of the nose as a whole. Secondly, variations of the conductivity in time may be observed easily, and this may be even a more valuable data than only one observation because of the wellknown nasal cycle (Keuning, 1963).

Thirdly, the open tube for the flow measurement adds hardly any resistance to the nose and so the breathing may stay within physiological limits. There is only one disadvantage in this type of measurement. That is the application of the measuring tubes to the nostrils. By this the valve action is partly suppressed. This can be besided by using a mask over the nose and connecting the flow-tube to this mask. The pressure has then to be measured by putting the pressure-tube in the mouth, but this latter has its own disadvantages.

SUMMARY

In this paper a description is given of an apparatus which measures directly the nasal conductivity. The nasal conductivity measured is defined as the discharged air volume is ccm per second at a pressure of one cm of water. The conductivity is recorderd on paper.

Actually the pressure and the square value of the flow are measured by means of strain-gauge pressure transducers and the apparatus has an electronic divider which performs the division $\frac{\dot{V}^2}{P}$.

The results of the measurements of nasal conductivities give a practical constant value of the conductivity during the greater part of the breathing cycle as well for inspiration as for expiration. This gives good evidence for the theory that during the greater part of the breathing cycle there is a square law relationship between pressure and flow.

Another evidence for this theory is given by making X-Y recordings of the square value of the flow and the value of the pressure during breathing on the screen of an oscilloscope. From these recordings one can see that during normal breathing only at the onset and the end of the inspiration or expiration there is a more linear relationship between pressure and flow. During the greater part there is a square law relationship.

RÉSUMÉ

Une nouvelle méthode pour mesurer la perméabilité nasale

On donne dans cette article la description d'un appareil mesurant directement la perméabilité nasale. La perméabilité se définit comme étant le débit,

exprimé en cm^3 par seconde sous une perte de charge d'1 cm d'eau. La perméabilité nasale est enregistrée sur papier. En fait c'est le carré du débit que l'on mesure, et l'appareil est pourvu d'un diviseur électronique, qui effectue la division $\frac{\dot{V}^2}{P}$.

Les résultats des mesures donnent une valeur constante pratique de la perméabilité nasale pendant la plus grande partie du cycle respiratoire, tant pour l'inspiration que pour l'expiration.

Ceci est un appui sérieux à la théorie qui veut qu'il existe, pendant la plus grande partie du cycle respiratoire, une relation au carré entre la perte de charge et le débit.

Une autre indication en faveur de cette théorie est donné par l'enregistrement d'un diagramme X-Y du carré du débit et de la valeur de la perte de charge sur l'écran d'un oscilloscope. On peut déduire de ces enregistrements, qu'au cours d'un cycle respiratoire normal, un rapport linéaire entre la perte de charge et le débit n'est pas valable que pour le début et pour la fin d'une inspiration ou d'une expiration, mais pendant la plus grande partie du cycle c'est une relation au carré qui s'applique.

REFERENCES

Cottle, Maurice H., a.o.: Rhino-sphygmomanometry and Rhino-revma-sphygmomanometry, Intern. Rhinology, **1**, 23, 1963.

Drettner, B.: Vascular reactions of human mucosa on exposure to cold, Acta oto-laryng., Suppl. 166, 1961.

Emms, E. T.: A novel single transistor RC oscillator, Electronic Engng., **32**, 506, 1960.

Keuning, J.: Rhythmic conchal volume changes, Intern. Rhinology, **1**, 57, 1963.

Malcomson, K. G.: The vasomotor activities of the nasal mucous membrane, J. Laryng., **73**, 73, 1959.

Seebohm, P. M. and Hamilton, W. K.: A method for measuring nasal resistance without nasal instrumentation, J. Allergy, **29**, 56, 1958.

Semerak, A.: Objektive Beurteilung der Nasendurchgängigkeit, Z. Laryng., **37**, 248, 1958.

Smith, C. Holt: Multipliers and dividers in A. C. computers, Electronic Engng., **32**, 714, 1960.

Spoor, A.: Aerodynamics, Intern. Rhinology, **1**, 19, 1963.

Stoksted, P.: Measurements of resistance in the nose during respiration at rest, Acta oto-laryng., Suppl. **109**, 143, 1953.

Tonndorf, J.: A note on the measurement of nasal flow resistance. Ann. Otol. Rhin. and Laryng., **67**, 984, 1958.

Wright, M. J.: A. Transistor phase sensitive demodulator of high performance, Electronic Engng., **34**, 698, 1962.

A. Spoor,
E.N.T. Department,
(Director: Prof. Dr H. A. E. van Dishoeck),
Academisch Ziekenhuis,
Leiden, Netherlands.