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Computer-controlled mechanical lung model for application in pulmonary function studies

A. F. M. Verbraak[1] J. E. W. Beneken[2] J. M. Bogaard[1] A. Versprille[1]

[1] Department of Pulmonary Diseases, University Hospital Dijkzigt, Dr. Molewaterplein 40, 3015 GD Rotterdam, The Netherlands

[2] Division of Medical Electrical Engineering, Eindhoven University of Technology, The Netherlands

Abstract—*A computer controlled mechanical lung model has been developed for testing lung function equipment, validation of computer programs and simulation of impaired pulmonary mechanics. The construction, function and some applications are described. The physical model is constructed from two bellows and a pipe system representing the alveolar lung compartments of both lungs and airways, respectively. The bellows are surrounded by water simulating pleural and interstitial space. Volume changes of the bellows are accomplished via the fluid by a piston. The piston is driven by a servo-controlled electrical motor whose input is generated by a microcomputer. A wide range of breathing patterns can be simulated. The pipe system representing the trachea connects both bellows to the ambient air and is provided with exchangeable parts with known resistance. A compressible element (CE) can be inserted into the pipe system. The fluid-filled space around the CE is connected with the water compartment around the bellows; The CE is made from a stretched Penrose drain. The outlet of the pipe system can be interrupted at the command of an external microcomputer system. An automatic sequence of measurements can be programmed and is executed without the interaction of a technician.*

Keywords—*Lung model, Mechanical, Simulation,*

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1 Introduction

THE COMPLEXITY of lung function measurements and data analysis entails the need for validation of all phases of data acquisition and analysis. Validation of software can be performed by the generation of raw data from a mathematical model with known parameters. When these raw data are read into a program to be tested, the calculated results must be in agreement with the parameters used for the simulation (VERBRAAK *et al.*, 1991). By means of a mechanical lung model with known physical characteristics, the reliability of the measurement equipment, as well as the software, can be tested. In addition, a lung model allows accurate reproduction of pulmonary function signals when required.

Many models have been described that could simulate only one type of breathing (PEDERSEN *et al.*, 1983; BOUHUYS and VIRGULTO, 1978; BOUTELLIER *et al.*, 1981). The model of Pedersen *et al.* consisted of a pressure chamber full of copper chips and was activated by the release of air from a container under pressure. After inflation to twice the atmospheric pressure, it delivered 8·2 l of air during deflation. Bouhuys and Virgulto developed a device to simulate flow-volume curves by means of a cam mounted on a motor. By changing the cam, the shape of the flow-volume curve could be altered. Maximum flow could be changed by changing the size of an orifice in the outlet line. Boutellier *et al.* devised a piston-pump system to generate respiratory data. This model was able to generate breaths of constant volume and known gas concentrations. With all the above-mentioned systems, different types of breathing could be simulated by a manual change of the mechanical hardware, whereafter the simulation had to be restarted.

The introduction of computers has made it possible to change quickly between different simulator settings. Meyer designed a microcomputer-controlled respiratory servo system, using a hydraulically operated cylinder-piston and solenoid valve assembly (MEYER, 1983). The flexibility in selecting different breathing patterns allowed the implementation of complex sequences of breathing manoeuvers. Myojo described a breathing simulator with a split/cam valve without a piston/cylinder or bellows (MYOJO, 1989). The opening of the split/cam valve was controlled by a stepper motor under microcomputer control. This system allowed inspiratory flow patterns to be simulated as seen during spontaneous breathing, although it was not reported whether it was possible to change breathing patterns during simulation. Jansen *et al.* described a computerised ventilator system for animal studies (JANSEN *et al.*, 1989). Instead of the hydraulic system of Meyer, they used an electromechanical system, which was relatively small and easy to implement in the construction of a ventilator. The system

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Correspondence should be addressed to A. F. M. Verbraak.

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was able to perform special breathing manoeuvers inserted between breathing cycles of normal mechanical ventilation.

We have adapted the idea of computer control to drive a physical lung model for use in the pulmonary function laboratory to test lung function equipment, to validate software, to simulate impaired pulmonary mechanics, and to perform simulations of special breathing patterns and pulmonary pathology. Primarily, we intend to use this computer-controlled mechanical lung model (CCML) to perform pulmonary function tests on special breathing patterns in a body plethysmograph (BP). We describe the mechanical lung model and some examples of its applications.

2 Mechanical lung model

The lung model consists of three components: the actual lung model, an electromechanical servo system and a computer system to control the model.

2.1 *Mechanical construction*

The construction of the mechanical lung model is shown in Fig. 1. It consists of a round container made from glass, an aluminium upper plate and a piston at the bottom. The container is filled with water, which represents pleural space, and is called the liquid-filled compartment (*LFC*). It contains two concertina bellows V_1 and V_2 that represent the two lung compartments. The *LFC* contains two connections P_{liq} and *LFC–CS*. P_{liq} is used to fill the *LFC* with water and to measure the pressure inside the *LFC*, and *LFC–CS* is used to connect the water compartment around the compressible element (see below) with the *LFC*. Leakage between the piston and container is prevented by other bellows. Vertical movements of the piston are established with an electrical motor, where rotation is transformed to longitudinal displacement by precision satellite roller screws*

To simulate breathing patterns, the piston is driven by a steering program according to a predefined pattern. The position of the piston is sensed by a potentiometer P_{piston}, attached to the piston. Two optical switches S_1 and S_2 at the two extreme positions of the piston are activated when the piston reaches its end limits. When one of the optical switches is activated, the system is stopped. The lung bellows have a conducting rod T_1 and T_2, respectively, which is attached to an aluminium bottom plate. A potentiometer is attached to each rod to sense the position of each bellows separately. The bellows V_1 and V_2 have outlets A_{w1} and A_{w2}, respectively, which give access to the 'airways', and two outlets P_{A1} and P_{A2} to measure the pressure inside the bellows.

As water is incompressible, the volume displacement of the piston is equal to the sum of the volume changes of the two bellows. Thus the position of one bellows depends on the position of the piston and that of the other bellows. As the diameter of the bellows is smaller than the diameter of the piston, the bottom plates of the bellows move faster than the piston. This difference in motion is even larger when one of the bellows is fixed or restricted in movement. Therefore, without precautions, a collision can occur between piston and bellows, and between bellows and the upper place, depending on the amount of water in the *LFC*.

The airways are simulated by perspex pipes, which have been constructed in a modular way. Flow resistances and flow and pressure transducers can be positioned at different places. A standard Lilly pneumotachometer head† fits into the pipes.

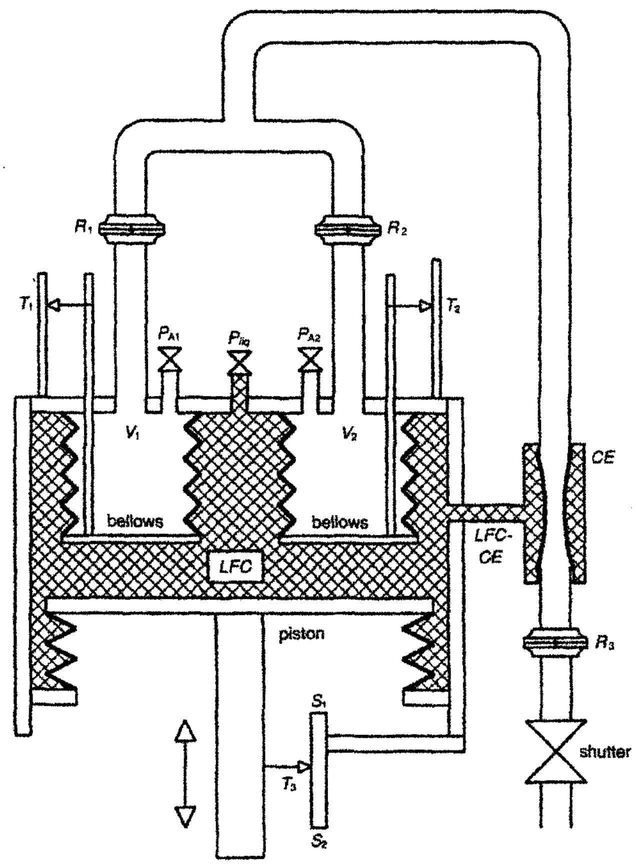


Fig. 1. *Mechanical lung model; airway segments R_1 and R_2 are connected to the outlets of concertina bellows; P_{liq} is the outlet to measure the pressure of LFC; liquid in LFC can be connected to the liquid around CS through the connection LFC-CS; S_1 and S_2 are optical switches for end limits; T_1 and T_2 are potentiometers for position of bellows; T_3 is potentiometer for position of piston; P_{A1} and P_{A2} are outlets to measure pressure inside bellows*

A compressible element, made of a stretched Penrose drain, can be inserted in series with the airway elements. The Penrose drain, a thin-walled floppy rubber tube, is fixed in a vertical position between the two pipes (Fig. 1) and stretched by about 10%. The pipes have an external diameter of 2×10^{-2} m. The vertical position of the Penrose drain can be shifted over 4×10^{-2}. The maximum length of the drain is 10^{-1} m. The drain is surrounded by water and is in direct contact with the *LFC*. The compressible element simulates the changing resistance in the airways due to the changing transmural pressures present in the physiological situation. At the outlet/inlet of the pipe system, a shutter can be used to close the airway opening. During occlusion of the pipe system, very high positive and negative pressures can develop in the *LFC* and bellows due to the power of the motor-driven system. Therefore, a protective device is constructed in the pipe system to avoid the destruction of the pressure transducers or bellows.

2.2 *Bellows*

The rubber bellows are stabilised by metal rings inside and outside each fold. The compliance of the bellows is determined when the bellows are outside the lung model and is about 20 L kP^{-1} for each. The bellows in the lung model are subject to forces caused not only by the compliance of the bellows itself, but also be its weight and that of the attached materials, the pressure inside the bellows and the pressure of the water

* Rollvis, Switzerland
† Jaeger, Germany

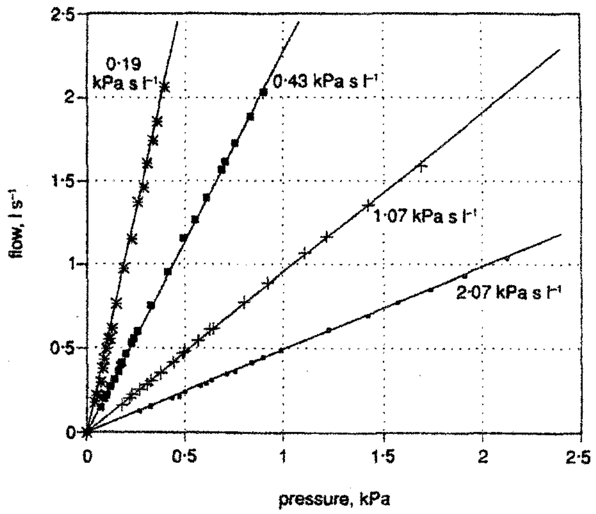


Fig. 2 *Relationship of flow and pressure for resistance elements*

around the bellows. The influence of the pressure against the side of the bellows is neglected. The diameter of the bottom plate of the bellows is 0·15 m. The position of the bottom of the bellows can be derived from the volume of the bellows, according to

$$h_{b,1} = \frac{V_{A,1}}{A} \times 10^{-3} \tag{1}$$

where A = effective surface of the bottom plate of each bellows, in m^2; $h_{b,1}$ = position of the bottom plate of bellows 1 (lung 1), in m; and $V_{A,1}$ = volume of the bellows without external pipes, in l.

The level $h = 0$ is the level of the bottom plate when $V_{A,1}$ is zero. The pressure at level $h_{b,1}$ in the water of the *LFC* near the bottom of the bellows is equal to the pressure at all other points h in the *LFC* if a correction for the hydrostatic pressure component is made. This relationship is given by

$$P(h) = P(h_{b,1}) - \rho \times g \times (h - h_{b,1}) \tag{2}$$

where $P(h)$ = pressure, in kPa, in the water at level h; $P(h_{b,1})$ = pressure, in kPa, in the water at level $h_{b,1}$; ρ = density of the water, in kg m^{-3}; and g = constant of gravitation, in m s^{-2}.

For the stationary situation, the pressure from the bellows on the water is equal to the pressure from the water on the bellows. This pressure is the summation of the pressure due to the elasticity of the bellows as a function of the position, the pressure inside the bellows and an offset pressure. This can be presented as

$$P(h_{b,1}) = \frac{k}{A} \times h_{b,1} + P_{A,1} + P_{off} \tag{3}$$

where P_{off} = pressure, in kPa, at $h_b = 0$, when $P_A = 0$ and $P_{grav} = 0$; $P_{A,1}$ = pressure in the bellows 1, in kPa; and k = elastic constant of the bellows at position $h_{b,1}$ in N m^{-1}.

The offset pressure is caused by the level of measurement, an additional pressure component due to the mass of bellows and attached materials, and a pressure due to the elasticity when level h is zero. The compliance is defined as $C = \Delta V_A / \Delta P(h_{b,1})$, giving

$$\frac{1}{C} = \frac{\Delta P(h_{b,1})}{\Delta V_{A,1}} = \frac{\left(\frac{k}{A} + \rho \times g\right) \times \Delta h_{b,1}}{A \times \Delta h_{b,1}} = \left(\frac{k}{A^2} + \frac{\rho \times g}{A}\right) \tag{4}$$

where C = compliance for the bellows, in l kPa^{-1}.

In accordance with eq. 4, when the bellows are placed in the water, the compliance decreases to 1·6 l kPa^{-1}. When both bellows are functioning, a coupling exists via the water.

The minimal volume of the bellows and pipe system is 0·45 l. The height of the *LFC* is such that, when both lungs are used in parallel and filled with the proper amount of water, they cannot touch each other. When only one of the bellows is used, the piston and bellows can touch each other. As the diameter of the lung bellows is smaller than the diameter of the piston, the bellows descend faster than the piston. The maximum displacement when only one of the bellows is used is 0·9 l. When both bellows are used in parallel, the maximum total displacement of the bellows is 2·3 l. The volume of the bellows for a given position of the piston depends on the amount of water in the *LFC*. The lung model does not fulfil the isothermal conditions as present in the lungs (Bargeton and Barrès, 1969). Isothermal conditions mean constant temperature and can be achieved by full and direct heat exchange with the surroundings; the other extreme, adiabatic conditions mean no heat exchange. For the model, a partial heat exchange exists. According to Poisson's law, an exponential relationship exists between pressure and volume (PV^k = constant). Depending on the rate of change of pressure and volume, k is between 1·0 (isothermic) and 1·44 (adiabatic). Inclusion of a thermal buffer, e.g. thin copper slices to approximate the isothermic condition, is not possible because of the changing volume of the bellows. For a sine wave pattern with different frequencies, we measure this value k. For volume changes with a sine wave pattern of 1 Hz, this value is 1·2.

2.3 *Resistances*

The resistance elements are constructed according to the work of Mecklenburgh (Mecklenburgh, 1988). They consist of a stack of layers of stainless-steel mesh. Different resistances are obtained by altering the number of layers or type of mesh. The packed layers are fitted in a Lilly pneumotachograph head. In this way, four resistance packages are made: 2·07, 1·07, 0·43 and 0·19 kPa s l^{-1}. The resistance elements are highly linear, as shown in Fig. 2. The resistance and pressure relationship of the compressible element has to be measured each time its characteristics, such as tension, material and length, are changed.

2.4 *Electromechanical servo system*

The electromechanical servo system consists of a motor§, a servo-contoller* and a custom-made electronic control module. The servo-controller controls the power to the motor by high-frequency chopping of the DC voltage. The difference between the programmed position and the actual position of the piston is converted by the electronics to drive the piston to the programmed position. The actual position is measured by the potentiometers attached to the piston and bellows. The electronic control unit also performs a first check on the allowed position range of piston and bellows. If one of the switches is activated, further movement of the piston is blocked until the piston is directed by the computer in a reverse direction. The amplitude diagram is constant up to 0·7 Hz; the 1 dB point is at 2·25 Hz, and the 2 dB point is at 3·35 Hz. These values are satisfactory for our applications during normal resting breathing.

2.5 *Microcomputer and software*

The operation of the lung model is controlled by a microcomputer. A program reads a control file from disk, in

§ Mavilor, model 600
* Infranor, model 90/20

which the measurement procedure is defined. The most important items to be defined are the type of breathing pattern, the number of breaths to be repeated, the sequence of breathing patterns and control of digital output. The program generates the analogue signal that steers the piston to a defined position. To simulate breathing patterns, programmed signals and measured signals from patients can be used. The software also reads some digital input to detect the state of the valves of the lung function equipment, for example. Depending on the values of these imputs, the steering program waits, restarts or ends its operation. The controlling software can be imbedded in a larger program to enable the microcomputer to perform a sequence of application programs.

During operation, the program calculates the next position of the piston and compares its consequences with the permitted range of positions of piston and bellows. If correct, the lung model is steered to the new position; otherwise, the lung model is held in the same position and waits for the next defined position, or is stopped by the operator. At certain moments, defined in a control file, digital output can be activated during the measurements.

2.6 *Calibration of lung model*

Calibration implies the definition of the relationship between the steering signal and the position of the piston, and the relationship between the signal of the displacement transducers and the position of piston and bellows. Through these relationships, at each moment the distance between each bellows and piston and that between the extreme positions are known. A recalibration has to be performed if the amount of water is changed.

2.7 *Transducers*

The pressures at outlets P_{liq}, P_{A1} and P_{A2} are measured by pressure transducers†, flow is measured by a pneumotachometer§ and volume changes are measured by a spirometer**.

3 Applications

The flexibility of the CCML can be illustrated by two applications.

3.1 *Compliance and airway resistance*

Lung compliance is the ratio between volume change (ΔV) and the simultaneous change in transpulmonary pressure P_{tp} (MURPHY and ENGEL, 1978). In routine lung function testing, P_{tp} is measured as oesophageal pressure P_{es} minus mouth pressure P_{mouth}. In patients, the P_{es} is measured by means of a latex balloon inserted in the oesophagus. If the compliance is estimated when volume and P_{es} are continuously changing, the compliance is called dynamic. If a patient expires slowly after a deep inspiration, the compliance estimate is called quasi-static.

When the patient's compliance is constant and the pressure drop in the bronchial airways is linear with flow V', the relationship between P_{es} and V' can be described as

$$P_{es} = P_{es,\,off} + R \times V' + \frac{\int V'\,dt}{C_L} \tag{5}$$

where $P_{es,\,off}$ = pressure offset, in kPa; R = flow resistance, in kPa s l^{-1}; V' = flow, in l s^{-1}; and C_L = constant compliance, in l kPa^{-1}.

† Valydyne, USA
§ Jaeger, Germany
** Lode, Holland

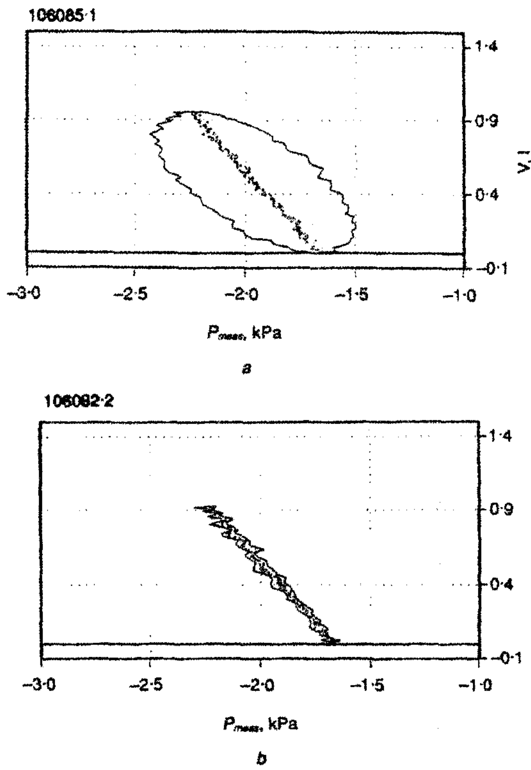


Fig. 3 *(a) P_{meas} is pressure difference between P_{liq} and ambient pressure P_{bar} and reflects pressure drop over liquid, bellows and additional resistance R_1; (b) P_{meas} is pressure difference between P_{liq} and the pressure inside bellows P_{A1}, which reflects pressure over bellows leaving resistance element R_1 out of the loop*

If flow is almost zero during a volume change, eq. 5 gives the quasi-static compliance. During cyclic changes in the volume, the dynamic compliance can be derived from this relationship.

The procedures to measure quasi-static and dynamic compliance and airway resistance can be simulated with the lung model. We use such simulations to test a software program developed for data acquisition in compliance measurements. Fig. 3 shows two examples of a $P_L - V$ relationship during simulated tidal breathing, for the situation with one lung bellows in use and without the compressible segment. The volume changes at the outlet are measured by means of a spirometer or by integration of flow. The pressure in the water P_{meas} is measured at outlet P_{liq}. In the lung model P_{meas} represents the oesophageal pressure P_{es}, as measured in patients.

Fig. 3*a* and 3*b* give the measurement values for the situation with one of the bellows fixed in the upper position (no displacement), and the other bellows connected with resistance R_1 and without the compressible element. For the measured curves, as given in Fig. 3*a*, the pressure difference is measured between P_{liq} and the pressure at the outlet of resistance R_1, P_{bar}, whereas Fig. 3*b* depicts the pressure as measured between P_{liq} and P_{A1}. In the first situation, P_{meas} gives the pressure drop over liquid, bellows and resistance, whereas, in the second situation, P_{meas} gives the pressure drop over the liquid and bellows without the pressure drop over the resistance element. Thus, the additional resistance can be calculated from the difference between the two measurements.

The continuous line in Fig. 3 shows the relationship between P_{meas} and volume during one cycle (inspiration and expiration). The dotted line shows the values for P_{es} when P_{meas} is corrected for the pressure drop over the resistance element. This pressure drop is calculated as the effective resistance

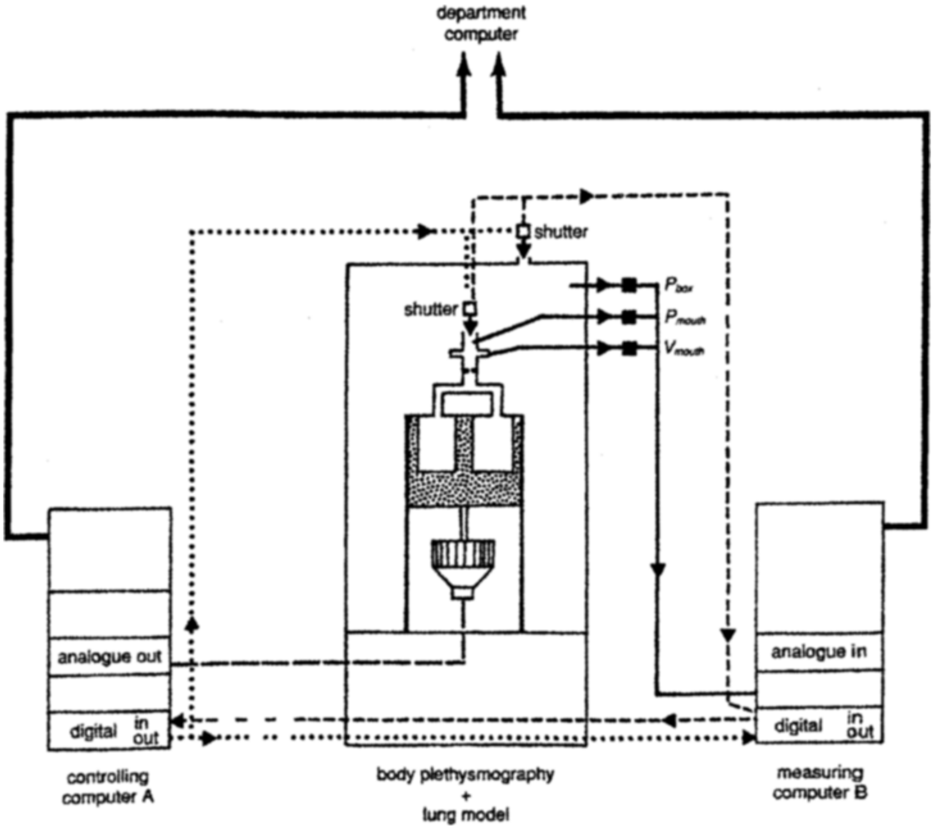


Fig. 4 *Lung model placed inside body plethysmograph; model and body plethysmograph are controlled by computer A; computer B performs the real-time measurements; P_{box} is measured pressure in box; P_{model} and V'_{model} are pressure and flow, respectively, at the combined outlet of both bellows*

$R_{L,eff}$ according to eq. 6. The irregularities on the curves are due to movement resistance (stick-slip) of the conducting rods. First, the resistance $R_{L,eff}$ is calculated as follows:

$$R_{L,eff} = \frac{\oint P_{es}\, dV}{\oint V'\, dV} \qquad (6)$$

It has previsouly been demonstated (HOLLAND *et al.*, 1986) that the effective resistance is valid measure for the resistance over the whole loop. With this resistance, each measured point P_{meas} is corrected for pressure drop due to the airflow, whereafter P_{off} and (from the slope of regression lines through the corrected points) C_{eff} are calculated. The values calculated with these methods are presented in Table 1 for the situation where pressure is measured over the bellows and additional resistance (Table 1*a*) and where pressure is measured over the bellows alone (Table 1*b*). The resistance values of the inserted resistance elements are given (column 2). The influence of bellows and water on these values ('null-resistance') can be calculated from Table 1*a* when no extra resistance is added (measurement 1), or from Table 1*b* when the pressure is measured between the *LFC* and the air in the bellows. The calculated resistances from Table 1 are corrected for the null resistance, which is the first value of each row for the calculated resistances. This results in the corrected resistance value $\Delta R_{eff,cor}$ given in Table 1. These corrected resistance value have a discrepancy of approximately 3%. The value for the compliance in Table 1*a* decreases with increasing resistance, whereas in Table 1*b* this values remains almost constant at approximately $1{\cdot}64\,\mathrm{l\,kPa^{-1}}$.

3.2 *Body plethysmography*

With the lung model, body plethysmographic (*BP*)†† measurements can be simulated; for this the lung model has to be placed in the body plethysmograph. With the *BP* (DUBOIS *et al.*, 1956), which has become a standard technique in many clinical and research laboratories, the thoracic gas volume (*TGV*) and airway resistance can be determined. The *BP* is a rigid box in which a volunteer or patient is seated surrounded by air. When a volume change occurs in a patient by compression or expansion of air ΔV_C, the volume of air in the box outside the patient ΔV_{bx} changes in the opposite way ($\Delta V_C = -\Delta V_{bx}$), which results in a pressure change ΔP_{bx} (Boyle–Gay Lussac). The relationship between ΔP_{bx} and ΔV_{bx} is found by calibration of ΔP_{bx} against imposed volume changes of air in the closed box. ΔV_C can then be found by measuring ΔP_{bx} in the *BP*. Of course, the volume of the air in the box, when a patient is seated in the *BP*, has to be corrected for the patient's volume, which is taken equal to body-weight in kg. Calculated ΔV_C is equal to compressed alveolar volume when influences of other closed air spaces inside the patient, such as air in the abdomen, are neglected.

In general, two types of measurement are performed. Phase A involves breathing against a shutter. *TGV* can then be calculated from the relationship between the pressure changes at the mouth, which reflect the pressure changes inside the lung (no flow situation), and the volume changes of the lung. Phase B occurs during tidal breathing, when the lung volume of the patient is known from the procedure described in phase A, and the alveolar pressure changes ΔP_A can be derived from ΔP_{bx}. When there is more than one lung compartment, the measured ΔP_A is the volume weighted mean of the pressures in the different compartments. The slope of the relationship between P_A and flow at the mouth (V'_m) reflects the flow resistance of the airways.

Fig. 4 shows the situation with the mechanical lung model placed in the *BP*. When the lung model is used, control of the *BP* is by the computer, which steers the lung model. The hand control of the *BP* is extended with a program selection using

†† Jaeger, Bodytest

Table 1 (a) measurements 1–5 are for the set-up in Figs. 3a and b, when pressure P_{meas} is measured between the liquid and outlet of the lung model; (b) measurements 6–10 are for the set-up in Fig. 3b, where pressure P_{meas} is measured between the liquid and air inside the bellows

	measurement number	implemented resistance, $\mathrm{kPa\,s\,l^{-1}}$ R_1	stoke volume, l V	(SD)	calculated compliance, $\mathrm{l\,kPa^{-1}}$ C_{eff}	(SD)	calculated resistance, $\mathrm{kPa\,s\,l^{-1}}$ R_{eff}	(SD)	corrected resistance, $\mathrm{kPa\,s\,l^{-1}}$ $R_{eff,\,cor}$	(Δ %)
	1	0·00	0·909	(0·004)	1·666	(0·012)	0·057	(0·001)	0·000	
	2	0·19	0·909	(0·004)	1·634	(0·014)	0·252	(0·001)	0·195	(3%)
a	3	0·43	0·902	(0·004)	1·585	(0·010)	0·501	(0·003)	0·444	(3%)
	4	1·07	0·897	(0·002)	1·458	(0·025)	1·159	(0·004)	1·102	(3%)
	5	2·07	0·877	(0·003)	1·390	(0·051)	2·249	(0·009)	2·192	(6%)
	6	0·00	0·914	(0·003)	1·653	(0·010)	0·010	(0·001)	0·000	
	7	0·19	0·908	(0·002)	1·641	(0·008)	0·011	(0·001)	0·001	
b	8	0·43	0·904	(0·002)	1·636	(0·009)	0·013	(0·001)	0·003	
	9	1·07	0·899	(0·004)	1·628	(0·006)	0·017	(0·000)	0·007	
	10	2·07	0·874	(0·004)	1·621	(0·012)	0·023	(0·001)	0·013	

The second column gives the values for the resistance elements. The stroke volume, the calculated values for C_{eff} and R_{eff} are given, together with the standard deviation (SD). The column $\Delta R_{eff,\,cor}$ is the calculated resistance value R_{eff} of each row diminished by the first element in the same row, where no additional resistance is implemented (e.g. 0·195 is 0·252 − 0·057). The last column for measurements 1–5 gives the percentage deviation between implemented resistance and $\Delta R_{eff,\,cor}$.

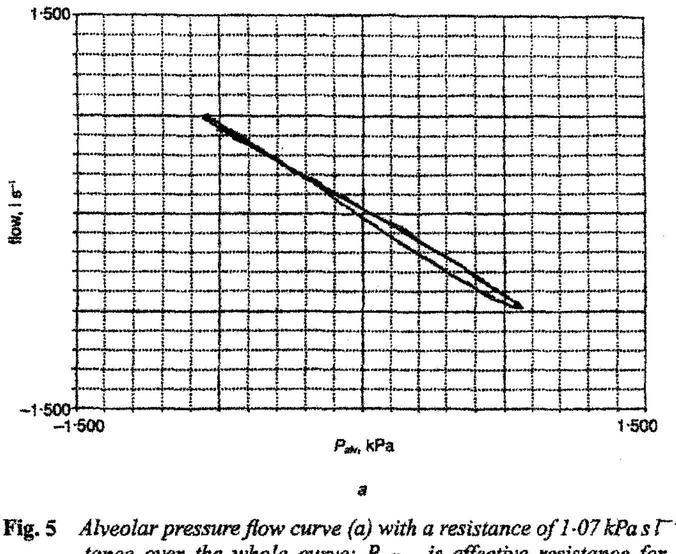
external control lines. This was done in two ways first, by simulating the circuit interruption of the push-buttons and secondly, by direct control of the valves. The 'controlling' microcomputer steers the model and sets the digital output for the lung function equipment to select the different phases of the measurement. The direct connections between the 'controlling' and 'measuring' computers are also indicated in Fig. 4. These connections are optional and can be used in situations where the 'measuring' computer also has to perform signal analysis of the measurement, e.g. the 'measuring' computer holds up the 'controlling' computer until the measured signals are analysed. In this way, together with the developed software, several predefined measurements can be performed.

A typical example of calculated P_A and V'_m is given in Fig. 5a and b. In Fig. 5a, only an additional constant resistance of $1{\cdot}07\,\mathrm{kPa\,s\,l^{-1}}$ is attached to the lung model. The calculation of the effective resistance is performed according to eq. 6, where P_A was used instead of P_{es}. For the measurement of Fig. 5b, a compressible element is inserted that simulates patients with expiratory flow resistance. The inspiratory resistance $R_{eff,\,insp}$ remained the same ($1{\cdot}15$–$1{\cdot}06\,\mathrm{kPa\,s\,l^{-1}}$), whereas the resistance during the expiration part $R_{eff,\,ex}$ increased from 1·11 to $1{\cdot}88\,\mathrm{kPa\,s\,l^{-1}}$. $R_{eff,\,tot}$ is the resistance over inspiration and expiration together. The reproducibility is shown in Fig. 5, where at least three loops are plotted. The irregularities, seen in Fig. 5b, are due to the stick-slip of the rods.

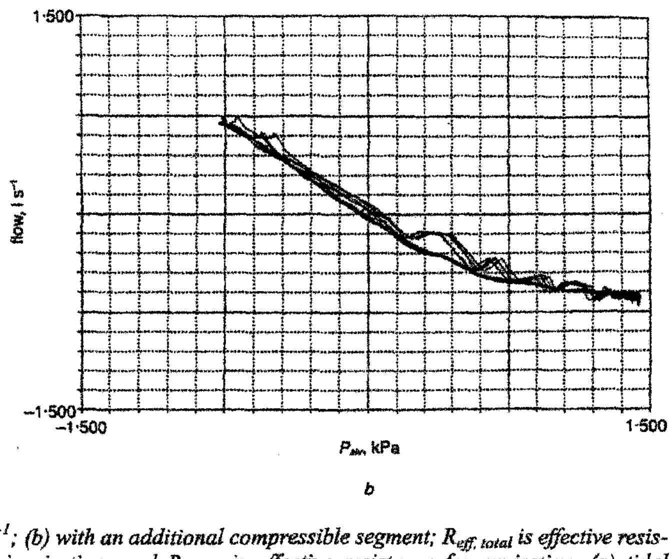
4 Discussion and conclusions

The computer-controlled mechanical lung model is used to test lung function equipment and to validate software. Owing to its construction, it can only be used as an active lung model, which means that it cannot be used to simulate an artificially ventilated lung.

The examples illustrate some of the possibilities of the mechanical lung model. Furthermore, it has been used to test other lung function measurements in the pulmonary laboratory, such as spirography, the closed-circuit helium-dilution method and pneumotachography; these are not discussed in this paper. All simulations had to be performed within a frequency and



a



b

Fig. 5 *Alveolar pressure flow curve (a) with a resistance of $1{\cdot}07\,\mathrm{kPa\,s\,l^{-1}}$; (b) with an additional compressible segment; $R_{eff,\,total}$ is effective resistance over the whole curve: $R_{eff,\,in}$ is effective resistance for inspiration; and $R_{eff,\,ex}$ is effective resistance for expiration. (a) tidal volume = 0·23 l; $R_{eff,\,insp}$ = $1{\cdot}15\,\mathrm{kPa\,s\,l^{-1}}$; $R_{eff,\,exp}$ = $1{\cdot}11\,\mathrm{kPa\,s\,l^{-1}}$; $R_{eff,\,total}$ = $\cdot13\,\mathrm{kPa\,s\,l^{-1}}$; (b) tidal volume = 0·22 l; $R_{eff,\,insp}$ = $1{\cdot}06\,\mathrm{kPa\,s\,l^{-1}}$; $R_{eff,\,exp}$ = $1{\cdot}88\,\mathrm{kPa\,s\,l^{-1}}$; $R_{eff,\,total}$ = $1{\cdot}46\,\mathrm{kPa\,s\,l^{-1}}$*

tital volume range restricted by the dynamical properties of the model.

Until now we have used the piston only as a flow generator. Therefore, during occlusion of the airway, the pressure inside the bellows can become very high. To cope with this problem, it is possible to measure the pressure inside the bellows and, using the known volume of the bellows, to calculate mean alveolar pressure. The software can be altered to steer the piston to follow a predefined pressure pattern and simulate body plethysmography with pressure signals as measured in patients. The software detects high pressures due to occlusion of the airways.

The *CCML* fulfils most of the demands. One problem with the model is the stick-slip ('stiction') of the rods, which causes irregularities in the signal, as shown in Fig. 3*a*, *b* and 5*b*, due to irregular movement of the bars through the guiding bearings. As can be seen for the simulation performed without the compressible element (Fig. 5*a*), the stick-slip is absent. In the situation shown in Fig. 5*a*, slight stick-slip results in a higher pressure in the water, which directly counteracts the effect of stick-slip on the movement of the lung bellows. In fig. 5*b*, the stick-slip is more pronounced due to the parallel compliance of compressible element and lung bellows. Therefore, a sudden pressure increase in the liquid is partially shunted to the surrounding of the compressible segment. Slight 'stiction' of the rod during expiration therefore results in a pressure increase in the water, causing a higher expiratory resistance by compression of the compressible segment. This pressure increase amplifies the primary effect of the stick-slip on the movement of the lung bellows. This higher pressure also causes an extra movement of the bellows that counteracts the stick-slip. As there is a complex time relationship between the effects of stick-slip and resistance of the compressible element, alveolar pressure disturbances occur that can be seen on measured alveolar pressures. An improvement may be obtained if the guiding bars are positioned in the centre of the bellows. The differences between implemented and calculated resistances are 2·6–5·8%. The calculated compliance in Table 1 is in the expected range.

A good quantification of the resistance and compliance values is an important aspect of the mechanical model when used with the parameter estimation techniques, as reported (VERBRAAK *et al.*, 1991). The change of the calculated compliance with high increased resistance elements must have its origin in the additional pressure components over the pipe system, because the value for the compliance, measured over the bellows alone, remains almost constant.

The amount of water that has to be displaced during operation causes a phase shift between the programmed volume displacement and the actual volume displacement of the piston. At high frequencies, this inertia of the system also results in a lower amplitude response. The compliance of each bellows cannot be changed yet. A change in compliance, for example, by additional springs, will be of value for extending the simulation possibilities of the model.

Although the model still has some limitations with respect to the mechanical construction and the limited flexibility in representing lung mechanics, it has been proven useful for testing equipment and software.

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References

BARGETON, D., and BARRÈS, G. (1969): 'Time characteristics and frequency response of body plethysmography,' *Prog. Respir. Res.*, **4**, pp. 2-23

BOUHUYS, A., and VIRGULTA, J. A. (1978): 'Calibration of flow-volume curves,' *Lung*, **155**, pp. 123–130

BOUTELLIER, U., GOMEZ, U., and MADER, G. (1981): 'A piston pump for respiratory simulation,' *J. Appl. Physiol.*, **50**,(3), pp. 663–664

DU BOIS, A. B., BOTELHO, S. Y., and COMROE, J. H. (1956): 'A new method for measuring airway resistance in man using a body plethysmograph; values in normal subjects and in patients with respiratory disease,' *J. Clin. Invest.*, **35**, pp. 327–335

HOLLAND, W. P. J., VERBRAAK, A. F. M., BOGAARD, J. M., and BOENDER, W. (1986): 'Effective airway resistance: a reliable variable form body plethysmography,' *Clin. Phys. Physiol. Meas.*, **7**,(4), pp. 319–331

JANSEN, J. R. C., HOORN, E., GOUDOEVER, J., and VERSPRILLE, A. (1989): 'A computerized respiratory system including test functions of lung and circulation,' *J. Appl. Physiol.*, **67**,(4), pp. 1687–1691

MECKLENBURGH, J. S. (1988): 'Construction of linear resistance units for a model lung,' *Med. Biol. Eng. Comput.*, **26**, pp. 552–554

MEYER, M. (1983): 'A versatile hydraulically operated respiratory servo system for ventilation and lung function testing,' *J. Appl. Physiol.*, **55**, pp. 1023–1030

MURPHY, B. G., and ENGEL, L. A. (1978): 'Models of the pressure-volume relationship of the human lung,' *Respir. Physiol.*, **32**, pp. 183–194

MYOJO, T. (1989): 'Breathing pattern simulation using slit/cam valve,' *Am. Ind. Hyg. Assoc. J.*, **50**,(5), pp. 240–244

PEDERSEN, O. F., NAERAA, N., LYAGER, S., HILBERG, C. and LARSEN, L. (1983): 'A device for evaluation of flow recording equipment,' *Bull. Eur. Physiopath. Resp.*, **19**, pp. 515–520

VERBRAAK, A. F. M., BOGAARD, J. M., BENEKEN, J. E. W., HOORN, E., and VERSPRILLE, A. (1991): 'Serial lung model for simulation and parameter estimation in body plethysmography,' *Med. Biol. Eng. Comput.*, **29**, pp. 309–317

5 Appendix: list of abbreviations and symbols

A_{w1}	= airway element for (lung 1)
A_{w2}	= airway element for (lung 2)
BP	= body plethysmograph
CCML	= computer-controlled mechanical lung model
CE	= compressible element
LFC	= water-filled compartment
V_1	= volume of concertina bellows (lung 1), in l
V_2	= volume of concertina bellows (lung 2), in l
P_{liq}	= Connection point to fill the *LFC* with water and to measure the pressure inside the *LFC*
LFC–CS	= connection between the *LFC* and the compressible element
P_{piston}	= position of the piston
S_1, S_2	= optical switches
T_1	= position of bellows 1
T_2	= position of bellows 2
A	= effective surface of the bottom plate of each bellows, in m_2
C	= compliance, in $\mathrm{l\,kPa^{-1}}$
C_{eff}	= compliance as calculated with R_{eff} in $\mathrm{l\,kPa^{-1}}$
C_L	= lung compliance, in $\mathrm{l\,kPa^{-1}}$
g	= constant of graviation, in $\mathrm{m\,s^{-2}}$
h_b	= position of bottom plate of a bellows, in *m*
k	= elastic constant of bellows at position h_b, in $\mathrm{N.m^{-1}}$
$P_{A,1}$	= pressure in the bellows (lung 1), in kPa
$P_{es,off}$	= pressure offset in the oseophagus, in kPa
$P(h)$	= pressure, in kPa, in the water at level h
$P(h_{b,1})$	= pressure (lung 1), in kPa, in the water at level h_b
P_{meas}	= measured pressure, in kPa

P_{off}	$=$ pressure, in kPa, at $h_b = 0$, when $P_A = 0$ and $P_{grav} = 0$	
ρ	$=$ density of the water, in kgm^{-3}	
R	$=$ flow resistance, in kPa s l^{-1}	
R_{eff}	$=$ effective flow resistance, in kPa s l^{-1}	
R	$=$ flow resistance, in kPa s l^{-1}	
$V_{A,1}$	$=$ volume of the bellows of lung 1 without external pipes and additional space, in l	
V'	$=$ flow, in l s^{-1}	
V'_m	$=$ flow at the mouth, in l s^{-1}	

Author's biography

A. F. M. Verbraak was born in Tilburg, The Netherlands in 1946. He received his MEE from Eindhoven University of Technology, Division of Medical Electrical Engineering, in 1976. He works in the Pulmonary Function Department of the University Hospital Rotterdam. His research interests are clinical physics and medical informatics in pulmonary diseases.