special communication

Interrupter airway and tissue resistance: errors caused by valve properties and respiratory system compliance

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Kessler, Volker, Georg Mols, Holger Bernhard, Christoph Haberthür, and Josef Guttmann. Interrupter airway and tissue resistance: errors caused by valve properties and respiratory system compliance. *J. Appl. Physiol.* 87(4): 1546–1554, 1999.—The interrupter technique is used to determine airway and tissue resistance. Their accuracy is influenced by the technical properties of the interrupter device and the compliance of the respiratory system. We investigated the influence of valve characteristics and respiratory system compliance on the accuracy of determining airway and tissue resistance by means of a computer simulation. With decreasing compliance we found increasing errors in both airway and tissue resistance determination of up to 34 and 71%, respectively. On this basis we developed a new occlusion valve, with special emphasis on rapid closing time and tightness in the closed state to improve the accuracy of resistance determination. The newly developed occlusion device greatly improves the accuracy of airway and tissue resistance determination. We conclude that respiratory system compliance is a limiting factor for the accuracy of the interrupter technique. To apply the interrupter technique in patients with extremely low respiratory system compliances, we need sophisticated technical devices.

computer simulation; initial pressure drop; interrupter technique; respiratory system mechanics

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AIRWAY RESISTANCE (Raw) and tissue resistance (Rti) can be determined by using the interrupter technique (5). If the flow at the airway opening is suddenly interrupted and zero flow is sustained for a few seconds, the airway pressure (Paw) curve shows three different phases: rapid change in Paw being a pressure drop in inspiration and a pressure rise in expiration (*phase 1*); damped oscillations of a second-order system usually lasting <30 ms (*phase 2*) (20, 28, 29); and slow change in Paw (*phase 3*). The initial rapid pressure change equals the resistive pressure drop across the natural and artificial airways (2), the damped oscillations reflect ringing of the inertive components of the respiratory system (20,

29), and the slow pressure change is caused by viscoelasticity and/or mechanical inhomogeneity of the lung (2) and, in the case of spontaneous breathing, by continued muscular effort.

The point of the Paw curve that marks the transition between *phase 1* and *phase 2* is called P1. The plateau pressure at the end of *phase 3* is called P2. P1 is the essential value for the determination of Raw, and the difference P1 − P2 is required to determine Rti. However, in the presence of pendelluft the pressure change from P1 to P2 does not reflect Rti exclusively, because it is not possible to separate tissue rheology from gas redistribution.

During an occlusion maneuver a certain amount of gas is still transmitted into or out of the lung. The amount of transmitted gas depends on the nonideal properties of the interrupter device, i.e., its finite closing time and its incomplete tightness. As a consequence, the pulmonary gas volume and the elastic recoil pressure change simultaneously. The amount of change in recoil pressure depends on respiratory system compliance. In effect, a long closing time and poor valve tightness, together with low respiratory system compliance, strongly influence the accuracy of the readings of the pressures P1 and P2.

The purposes of the present investigations are *1*) to perform a theoretical analysis of the influence of closing time and gas tightness of an interrupter device and of respiratory system compliance on the errors in determination of Raw and Rti; *2*) to screen the literature for the physical properties of the existing interrupter devices used thus far; *3*) to obtain specific design criteria for a new interrupter device; *4*) to develop a new interrupter device on the basis of these criteria; and *5*) to investigate the technical properties of the new interrupter device in the laboratory.

MATERIALS AND METHODS

Theoretical Analysis

The change in P1 and P2 because of the finite closing time and the incomplete tightness of the valve over a period of several seconds is dependent, first, on the additional volume

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ΔV passing the valve and, second, on the patient's respiratory system compliance Crs as described in general

$$\Delta P = \frac{1}{Crs} \cdot \Delta V \qquad (1)$$

where ΔP is change in pressure.

The additional volume ΔV passing the valve during its closing process was simulated in two ways: a worst case (wc) and a realistic case (rc) estimate. The ΔV for the worst case was calculated as the product of the full preocclusional flow and the simulated closing time. Depending on the actual respiratory system compliance, the additional postocclusional pressure change ΔP1 due to finite closing time can be calculated

$$\Delta P1_{wc} = \frac{1}{Crs} \cdot \Delta V_{wc} = \frac{1}{Crs} \cdot \dot{V}_{preoccl} \cdot t_{closing} \qquad (2)$$

where $\dot{V}_{preoccl}$ is preocclusional flow, and $t_{closing}$ is closing time.

Physically reasonable flow-closing profiles should generally behave like a one-quarter circle [e.g., Romero et al. (29)] or a one-half-bell-shaped function. To approximate comparable closing profiles, we assumed for the realistic case estimate a one-half-bell-shaped function, symmetrical to the point of one-half the preocclusional flow and one-half the closing time. Mathematically, the additional volume passing the valve in this case will be one-half the value of the worst case simulation. Therefore, the additional pressure change for the assumed realistic case will equal one-half the worst case estimate

$$\Delta P1_{rc} = \frac{1}{Crs} \cdot \Delta V_{rc} = \frac{1}{2} \cdot \Delta P1_{wc} \qquad (3)$$

The leakage volume ΔV passing the valve during an occlusion was calculated as the product of leakage flow ($\dot{V}_{leak}$) and occlusion time ($t_{occlusion}$), here simulated to be 5 s. Depending on respiratory system compliance, the additional slow pressure change ΔP2 due to leakage of the valve can also be calculated

$$\Delta P2 = \frac{1}{Crs} \cdot \Delta V = \frac{1}{Crs} \cdot \dot{V}_{leak} \cdot t_{occlusion} = \frac{1}{Crs} \cdot \dot{V}_{leak} \cdot 5\ s \qquad (4)$$

We investigated the influence of ΔP1 and ΔP2 on the accuracy of estimating Raw and Rti by means of a computer simulation. We simulated both the estimation of Raw with closing times of 5 and 15 ms and the estimation of Rti, assuming different leakage flows over an occlusion period of 5 s together with respiratory system compliances ranging from 80 to 10 ml/cmH$_2$O, thus simulating normal lungs and lungs with acute respiratory distress syndrome (ARDS).

To estimate the relative errors [ΔRaw (%) and ΔRti (%)], the additional pressure changes have to be related to reference values: we assumed Raw values of 2.5 and 4.5 cmH$_2$O·s·l^{-1} and Rti values of 2.5 and 7.0 cmH$_2$O·s·l^{-1} for normal anesthetized paralyzed subjects and for ARDS patients, respectively, as an average taken from data presented by Eissa et al. (11) and Tantucci et al. (33). These values do not represent individual or standard values but have been chosen as reference values in this study to estimate the error in the determination of Raw and Rti. Raw in ARDS patients is independent of flow, and, in normal anesthetized paralyzed subjects, it is reported to be flow dependent but to an extent that is negligible for our purposes. The values for Rti were selected at a flow rate of 1 l/s.

Equations 5 and *6* show the calculation steps for the estimation of ΔRaw (%)

$$\Delta Raw(\%) = 100 \cdot \frac{\Delta P1}{P_{init}} = 100 \cdot \frac{1/Crs \cdot \dot{V}_{closing} \cdot t_{closing}}{Raw \cdot \dot{V}_{preoccl}} \qquad (5)$$

with P_{init} being the initial pressure drop after rapid airway occlusion, and $\dot{V}_{closing}$ being the flow during the closing process.

On the assumption that the flow during the closing process is dependent on the preocclusional flow, as it occurs in both our simulation and reality, *Eq. 5* is simplified to give

$$\Delta Raw(\%) = 100 \cdot A \cdot \frac{1/Crs \cdot t_{closing}}{Raw} \qquad (6)$$

Factor *A* stands for the dependence on preocclusional flow; here, $A = 1$ for the worst case and $A = 0.5$ for the realistic case.

Equation 7 describes ΔRti (%)

$$\Delta Rti(\%) = 100 \cdot \frac{\Delta P2}{|P1 - P2|} = 100 \cdot \frac{1/Crs \cdot \dot{V}_{leak} \cdot t_{occlusion}}{Rti \cdot \dot{V}_{preoccl}} \qquad (7)$$

Occlusion Valve

The design of our newly developed occlusion valve is shown in Fig. 1. The occlusion valve consists of a conical valve cylinder made of Delrin (*E*) that is built in a valve case, with a conical valve seat made of brass (*F*) and a cone angle of $\alpha = 14.5°$. In the "valve-open" position, the boring of the valve
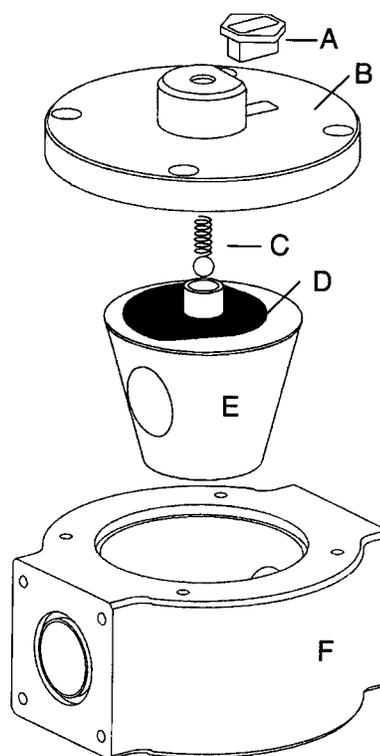

Fig. 1. Exposure of main parts of occlusion valve. *A*: infrared-emitter/sensor. *B*: *top* of valve case. *C*: spring and ball for downward movement (sealing) of rotating cylinder; *D*: black and white line for cylinder position measurements with infrared sensor. *E*: rotating conical-type cylinder with boring. *F*: *bottom* of valve case. All parts of driving unit (pneumatic drive) are not shown.

cylinder that is perpendicular to its axis of rotation coincides with the connection boring of the valve case. The inner diameter of the boring is 15 mm. The resistance of the valve in this position is <0.3 $cmH_2O \cdot s \cdot l^{-1}$ up to a flow rate of 2 l/s. To switch the occlusion valve into the "valve-sealed" position, the valve cylinder rotates 90° by using a pneumatic high-speed drive (type PRN1–90°, Sommer, Pforzheim, Germany) located under the valve case (not shown in Fig. 1). The operating cycle of valve closing is subdivided into two parts: *1*) the valve cylinder is lifted 0.1 mm without rotation by using a mechanical link motion (not shown) and the pneumatic drive is accelerated, and *2*) the valve cylinder is rotated 90°. Passing the 55° angle, the valve reaches the valve-closed position, at which point the boring of the rotating cylinder no longer has any visible contact with the connectors of the valve case. For most applications, this position can already be seen as gastight; any leakage is only allowed by the space between the lifted cylinder and valve case, 0.1 mm. During the last 5° of the cylinder's rotation, it is moved downward by the force exerted by the compression spring (*C*), which also seals the occlusion valve in the valve-sealed position. During the operating cycle, the position of the valve cylinder is continuously measured with an optical infrared emitter/sensor (*A*; type SFH 900, Siemens, Munich, Germany). The optical measurement of the actual cylinder position is realized by moving a special-shaped edge separating two different infrared-reflecting materials through the visual field of the sensor (*D*).

Further technical features are *1*) guaranteed automatic reopening of the valve in case of a breakdown of the gas and/or power supply; *2*) occlusions performed by means of a boring turning itself out of the gas stream, therefore not producing any artificial drafts that can cause artifacts in the pressure transducers, as is possible in the case of an occlusion plate turning inside a lumen of a tube, as described (e.g., Ref. 8); *3*) computer control: possibility of single or repetitive flow-and-volume-triggered occlusions throughout the whole breath, with occlusion duration freely selectable in 2-ms steps and a delay time between trigger signal and start of cylinder rotation of 17 ms; and *4*) dismantling and resterilization of all parts of the valve, with its physical characteristics remaining unaffected.

Laboratory Investigation

The experimental setup for determination of the valve characteristics includes a ventilator (Evita2, Dräger, Lübeck, Germany) to generate a constant flow rate ranging up to ±2 l/s, with the frequency set to 20 breaths/min and inspiration-to-expiration ratio set to 1:1. Gas flow rate was measured with a pneumotachograph (Fleisch no. 2, Metabo, Epalinges, Switzerland) connected to a differential pressure transducer (CPS1, Hoffrichter, Schwerin, Germany). The response of the pneumotachograph was linear over the experimental range of flow rates. The pressure drop over the valve was measured by using two relative pressure transducers (1210A, ICSensors, Milpitas, CA) at two identical pressure-measuring sites located symmetrically to the axis of rotation of the occlusion valve. The response time of the pressure sensors was 1 ms, the common mode-rejection ratio of the amplifiers was 115 dB. The flow signal, pressure signals, and output signal of the optical sensor were sampled at a rate of 250 Hz and digitized 12 bits wide (SDM863, Burr-Brown, Tucson, AZ) for subsequent numeric analysis (SparcStation 4, Sun Microsystems, Palo Alto, CA).

Starting from the valve-open position, the conical cylinder of the valve was turned manually in steps of 5°. In each (static) position of the valve cylinder, three different flow

rates (1, 1.5, and 2 l/s) were realized, and measurements were repeated five times. To prevent Venturi effects from affecting differential pressure measurements during the near-closed position of the valve, tubes were added on both sides of the valve.

For precise determination of closing time in relation to valve position $t(\alpha)$, the output voltage of the optical sensor was measured during the closing process and stored with a digital storage oscilloscope at a sample frequency of 20 kHz (Nicolet 490, Madison, WI). In addition, the output voltage of the optical sensor was measured while the cylinder of the valve was turned manually in steps of 2°. The function $t(\alpha)$ was finally calculated by using a second-order polynomial least squares fit.

To quantify the leakage of the valve in the valve-closed ($>55°$) and valve-sealed positions (90°), the ventilator (Evita2) was set in the continuous positive Paw mode, which ranged from 5 to 20 cmH_2O, while the whole valve was submerged underwater to enable the capture of the smallest volumes of gas leakage. The time during which a certain leakage volume passed the valve was measured, and the leakage flow was determined as the quotient of leakage volume and time. The values were corrected for the hydrostatic pressure.

RESULTS

Theoretical Analysis

Figure 2 shows the relative error in the estimation of Raw obtained in the first simulation. The diagram shows that the error in the estimation of Raw *1*) increases with increasing closing time and *2*) increases with decreasing respiratory system compliance. In particular, for a very low compliance, which simulates severe ARDS, the error increases to 17 and 34%, assuming a closing time of an occlusion device of 15 ms
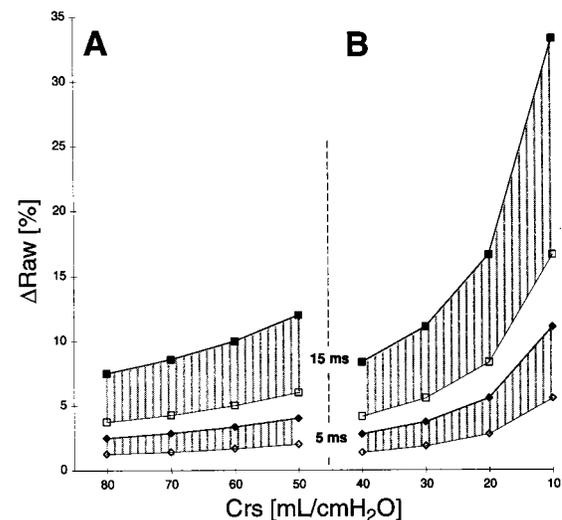
Fig. 2. Relative error in airway resistance [ΔRaw (%)], simulated by 2 different closing times of an occlusion device (5 and 15 ms), in relation to respiratory system compliance (Crs) in normal subjects (*A*; Raw = 2.5 $cmH_2O \cdot s \cdot l^{-1}$) and in patients with acute respiratory distress syndrome (ARDS; *B*; Raw = 4.5 $cmH_2O \cdot s \cdot l^{-1}$). Additional volume passing valve during closing process was estimated as worst case (solid symbols), calculated as product of full preocclusional flow and closing time, and realistic case (open symbols), on assumption of symmetrical one-half-bell-shaped decrease in flow over closing process. Results for most available occlusion devices would fall between each pair of lines (hatched area).

and realistic or worst case conditions, respectively. Even for very low closing times of 5 ms, the errors in the presence of extremely low respiratory system compliance are still as high as 6 and 12% for realistic and worst case estimates, respectively.

Figure 3 shows the relative error in the estimation of Rti obtained in the second simulation. The diagram also shows that the error in the estimation of Rti *1*) increases with increasing leakage flow and *2*) increases with decreasing respiratory system compliance. For a very low compliance and a leakage flow through the occlusion device of 10 ml/s over a period of 5 s, the error increases to 71%. With respiratory system compliances ranging from 80 to 20 ml/cmH$_2$O, only the use of extremely tight valves with leakage flows of ~0.5 ml/s produced errors that are generally <2%. For a very low compliance of 10 ml/cmH$_2$O, the error is still 3.6%.

Development of a New Occlusion Valve

Conductance during the closing cycle. Figure 4*A* shows the relationship between the conductance of our occlusion valve and the valve angle measured under static conditions. As the diagram shows, conductance is first markedly decreased, starting at a closing angle of 10°, resulting in a reduced closing time and an increase in trigger delay (between software command and actual begin of closing).

The relationship between time and valve position determined from the static voltage-angle measurements and the dynamic voltage-time measurements is mathematically described by a second-order polynomial least squares fit ($R^2 = 0.9997$)

$$t(\alpha) = -0.0015\alpha^2 + 0.3321\alpha + 0.086 \qquad (8)$$

As shown in Fig. 4*B*, the measured conductance-angle curve at a flow rate of 1 l/s has been transformed

Fig. 4. *A*: conductance (G) of new occlusion valve in relation to valve position during closing process (cylinder manually turned in steps of 5°), with flow rates ranging from 1 to 2 l/s. Values are given as means ± SD ($n = 5$ measurements). *B*: conductance-time curve of 1 closing cycle at preocclusional flow of 1 l/s, calculated from static conductance-angle curve and dynamic closing time-angle curve (*Eq. 8*). The curve is mathematically described by an exponential fit (thick line). Root mean square error of the fit is $0.08\ l \cdot s^{-1} \cdot cmH_2O^{-1}$.

into a function of time t by using *Eq. 8*. The transformed conductance-time curve was mathematically approximated by using an exponential fit

$$G = f(t) = G_0 \cdot e^{b(t-2)+c(t-2)^2} \qquad (9)$$

where G_0 is the conductance of the valve in the open position.

With $G(t) = G_0$ ($t < 2$ ms) and $b = c = -0.05$ ($t \geq 2$ ms), the root mean square error is $0.08\ l \cdot s^{-1} \cdot cmH_2O^{-1}$.

Leakage of the valve. Table 1 gives the leakage in the closed position (55–90°), which is <6 ml/s at a pressure difference over the closed valve of 10 cmH$_2$O. The valve remains in this status for ~4.5 ms, so that the leakage volume during this time would be <0.03 ml. Therefore,
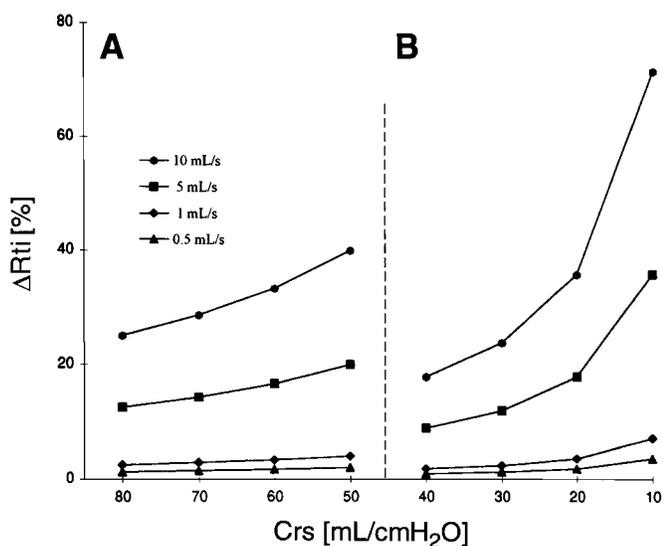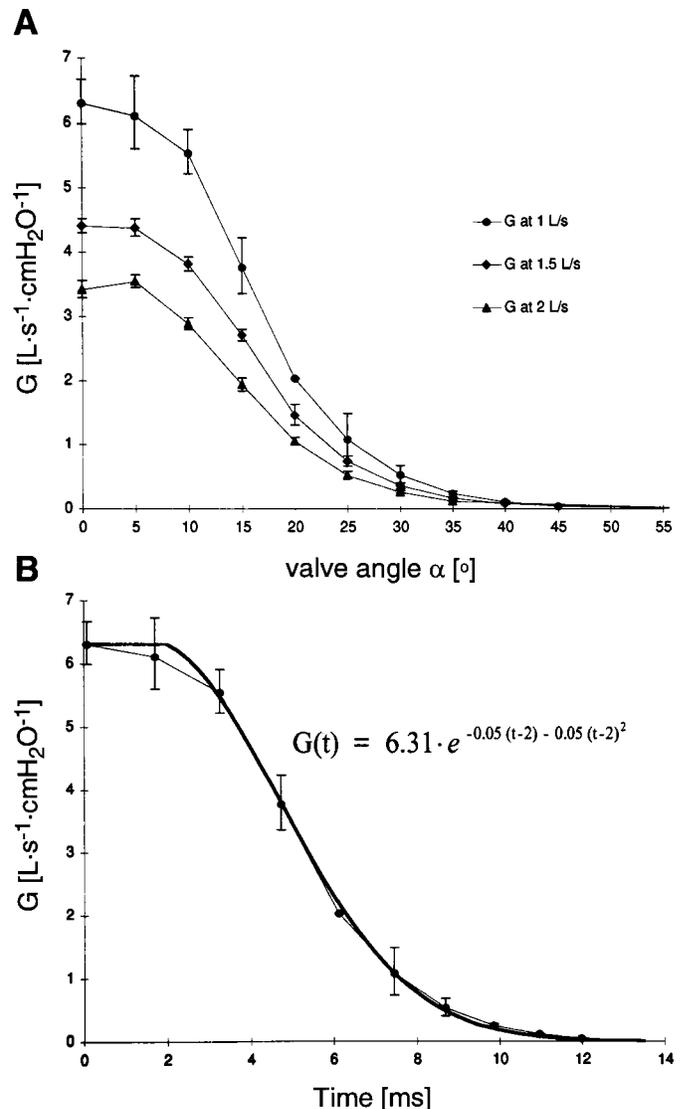
Fig. 3. Relative error in tissue resistance [ΔRti (%)] simulated by 4 different leakage flows, passing an occlusion valve over an occlusion period of 5 s, in relation to Crs in normal subjects (*A*; Rti = 2.5 cmH$_2$O·s·l^{-1}) and in patients with ARDS (*B*; Rti = 7.0 cmH$_2$O·s·l^{-1}). Reference values are taken at a flow of 1 l/s.

Table 1. *Leakage of the occlusion valve in the closed and sealed position*

Closed Position (55–90°) at $\Delta P = 10$ cmH$_2$O	
Valve angle, °	Leakage flow, ml/s
60	4.1 ± 0.9
70	5.6 ± 0.4
80	1.9 ± 0.1
Sealed Position (90°) at $\Delta P = 5$–20 cmH$_2$O	
ΔP, cmH$_2$O	Leakage flow, ml/s
5	0.11 ± 0.02
10	0.23 ± 0.02
15	0.36 ± 0.03
20	0.56 ± 0.07

Values are means $\pm$ SD ($n = 5$ measurements). ΔP, pressure difference over the valve.

the valve is effectively closed at a valve angle of 55°. Table 1 shows that the leakage in the sealed position (90°), with pressure differences over the sealed valve ranging from 5 to 20 cmH$_2$O, is <0.6 ml/s in all cases.

Total closing time is 17.5 ms. Because the valve is effectively closed at an angle of 55° and because conductance does not decrease until an angle of 10°, the effective closing time reduces to 10 ms. Therefore, the valve trigger delay increases by 3 ms, to 20 ms.

DISCUSSION

In the present study, we investigated the influence of respiratory system compliance and of the nonideal technical properties of an occlusion device on the accuracy of Raw and Rti determination. During an occlusion maneuver, a certain amount of gas is still transmitted through the occlusion device. The reasons for this gas transmission are mainly the finite closing time of the occlusion device but also its incomplete tightness in the closed state. As a consequence, the pulmonary gas volume changes, thereby inducing a concomitant change in alveolar pressure that is dependent on respiratory system compliance.

Especially in the presence of severely reduced respiratory system compliance, as seen in ARDS, the change in the elastic recoil pressure will be particularly high, and hence the errors in estimating Raw and Rti are large. However, for this particular group of patients a precise determination of respiratory system mechanics is urgently desired by clinicians at the bedside. For this reason, we developed a new occlusion device based on the theoretical evaluation of errors, eliminating most of the technical problems of the interrupter technique.

Theoretical Analysis

In the first computer simulation (Fig. 2), we found that the relative errors in Raw depended considerably on respiratory system compliance. With decreasing compliance the relative errors can become as large as 34%, caused by errors in the reading of P1.

The flow through the occlusion device during its closing process is bidirectional: into the lungs in inspiration (assuming the ventilator does not stop gas

delivery within 15 ms due to pressure limits) and out of the lungs in expiration, both resulting in a smaller initial pressure drop and consequently an underestimation of Raw (4).

P1 determination. Several methods for the correction of the errors in the determination of P1 have been described. To our knowledge Bates et al. (4) first investigated the influence of valve closing time on the estimation of respiratory system resistance. With their backextrapolation method, the authors could reduce the error in the estimation of Raw from 7 to 1–2%. However, because of the additional volume passing the valve during its closing process, the whole pressure-time curve is shifted to higher or lower values during inspiratory or expiratory occlusion maneuvers. The backextrapolation method would then only yield an under- or overestimated value for P1 (22).

A second correction method offers the use of the Fourier transform: on the assumption that the respiratory system is linear, the pressure signal can be expressed as the convolution of the flow signal with the impulse response of the respiratory system. The pressure signal can then be corrected as described by Romero et al. (29). However, this method requires knowledge of the flow profile during the closing process of the occlusion device.

Kochi et al. (22) first took account of the compliance of the respiratory system by mathematically correcting their resistance values. To use their correction method, however, one has to know the compliance of the respiratory system as well as the volume passing the valve during its closing process.

P2 determination. To our knowledge, the influence of the incomplete valve tightness on the accuracy of estimating Rti has not been investigated previously.

The second computer simulation revealed errors in the estimation of Rti of up to 71%, in the presence of a leakage flow of 10 ml/s and 5-s occlusion time and considerably decreased respiratory system compliance. The demand for a very tight valve becomes obvious: even with a leakage flow of only 0.5 ml/s, the relative errors in the determination of Rti rise to 3.6%, assuming a compliance of 10 ml/cmH$_2$O.

By using a correction method similar to that of Kochi et al. (22), these errors can be corrected

$$Rti^* = Rti \pm \frac{1/Crs \cdot \Delta V_{occlusion}}{\dot{V}_{preoccl}} \quad (10)$$

where Rti* is the true value of Rti and $\Delta V_{occlusion}$ is the change in occlusion volume. However, again, there are the same problems as mentioned above: determination of respiratory system compliance and determination of minimal leakage flows with conventional flow-measuring devices.

Generally, neither the flow during the closing process nor the leakage flow during an occlusion maneuver can be captured faithfully (29). The flow occurring during the closing process changes too quickly to be reliably measured, and the leakage flow during an occlusion is usually too small to be reliably measured with the same

flow-measuring device used for general flow measurements in humans. Consequently, the transmitted gas volume cannot be measured, and there is no easy way to estimate and to correct for the errors in the readings of P1 and P2 by using the method as described by Kochi et al. (22) and *Eq. 10*. The only way in which this would be possible is by analyzing the physical properties of an interrupter device and therefore knowing the closing functions and leakage flows under certain conditions. However, even then the correction possibilities are limited: with low respiratory system compliance the errors become very large, and to correct for errors of 50% or more is not reasonable. Therefore, the demand for a rapidly closing and sufficiently tight occlusion valve remains.

There is no influence of the finite closing time on the estimation of Rti, as reported by Bates et al. (4) and Sly and Bates (31). The problem, however, is the incorrect determination of P1 due either to the pressure oscillations, which occur immediately after occlusion, or to erroneous backextrapolation of the pressure signal. This results in an erroneous Raw as well as a false difference P2 − P1; therefore, Rti would also be incorrect. The precise determination of P1 is the most important point in using the interrupter technique to determine respiratory system resistances. One should put special emphasis on its precise determination as both Raw and Rti are dependent on the correct value of P1.

Neither simulation includes the case of normal respiratory system compliance and extremely increased resistance as seen in patients with chronic obstructive pulmonary disease. However, because of the increased time constant and the mechanical inhomogeneity, the use of the interrupter technique in patients with severe chronic obstructive pulmonary disease is controversial in any case (28).

Existing Occlusion Devices

Since the introduction of the airway occlusion technique by von Neergaard and Wirz in 1927 (24), many research groups have used this technique to determine respiratory system resistance in animals (17, 22), in excised isolated lungs (12), in patients during controlled mechanical ventilation (18), and in intubated spontaneously breathing patients during weaning from mechanical ventilation (27).

Many different types of interrupter devices have been used so far (see Table 2), and surely many of them have fulfilled the operating requirements. However, the physical properties of most devices were not described in detail. Therefore, for most devices it is unknown whether they can be used for fast short-term occlusions as well as for rapidly closing long-term occlusions. There was always the compromise between rapid closing on the one hand and long-term tightness on the other.

On the basis of these considerations and the results of our computer simulations, an occlusion device has to meet the following criteria: *1*) very short valve-closing time; *2*) strong tightness over a period of several seconds in the closed position; *3*) minimal resistance in the open position, especially for use in spontaneously breathing patients; *4*) small dead space, especially for the use in pediatrics; *5*) no vibrations or "turning drafts" because of the closing process to avoid additional artifacts; *6*) integrated measurement of the valve position to determine P1 on the time axis and to establish a closing function; *7*) maintenance of its characteristics under adverse day-to-day conditions and under continuous use; and *8*) minimal effect because of real-world difficulties, such as moisture accumulation or wear caused by repetitive cycling.

Unfortunately, the two most important demands, a fast closing time and tightness, are contradictory. Rapid closure in the physical sense usually means little friction between closing element and case, and little friction usually stands for poor tightness. This may be the reason that hardly any data are available in the literature (see Table 2) for both the closing time and the leakage of one valve.

New Occlusion Valve

As far as we know, there are no commercially available occlusion valves that satisfy all the necessary conditions mentioned above. Therefore, we have designed a new type of occlusion valve whereby valve closure and valve sealing are two physically separate processes. We constructed a valve that effectively closes within 10 ms and is pressure tight, with a minimal leakage flow in the sealed position of <0.6 ml/s at a pressure difference over the valve of up to 20 cmH$_2$O.

To estimate the additional volume passing the valve during its closing process, the flow has to be integrated to volume. We assumed the flow to be similar to the conductance profile, changing from $\dot{V}_0$ to zero according to *Eq. 9*

$$\dot{V} = f(t) = \dot{V}_0 \cdot e^{b(t-2)+c(t-2)^2} \qquad (11)$$

where $\dot{V}_0$ is the flow in the open position.

With $\dot{V}(t) = \dot{V}_0 = 1$ l/s ($t < 2$ ms) and $b = c = -0.05$ ($t \geq 2$ ms), the additional volume passing our valve during its 10-ms closing process is as small as 3.4 ml.

Where in the simulation is our new interrupter device positioned? An additional volume passing the valve of ~3.4 ml would place our valve between the worst case and realistic case simulations for a valve closing time of 5 ms at a preocclusional flow of 1 l/s (worst case: 5 ms·1 l/s = 5 ml; realistic case: 2.5 ml; see Fig. 2). The error in the estimation of Raw without any corrections for closing time would be generally <3%; only in patients with low respiratory system compliance values of 20 or 10 ml/cmH$_2$O would the error rise to 5 and 10%, respectively. However, these errors still appear to be very high but can finally be corrected further by using one of the correction methods for valve closing time described previously. A comparable error estimation in relation to respiratory system compliance has not been done by other authors so far, except by Kochi et al. (22) in cats.

Table 2. *Interrupter devices described in the literature*

Author, Yr (Ref. No.)	Application	Valve Type/Drive	Closing Time	Time Closed/ Duration Type	Type of Occlusion	Leakage	Dead Space
Clements et al., 1959 (10)	MV SB	Rotary valve Rotation at constant frequency	<50 ms	Equal intervals of unknown time	10 times/s		
Knudson et al., 1974 (21)	SB	Manual Electromagnetic	10 ms 30 ms	70 ms Opening shutter	Single Repetitive		
Schlesinger et al., 1980 (30)	SB, animals	Modified Ruben valve, reflex camera body	<60 ms	200 ms	Single	No (manually tested)	<8 ml
Clarke et al., 1982 (9)	SB	Two rotating disks Rotation at constant frequency	1 ms	100 ms	Repetitive (every 500 ms)		
Gottfried et al., 1985 (18)	MV	Pneumatical shutter (Rudolph 4200)*		100–200 ms	Single		70 ml
Bates et al., 1987 (4)	CS	Electromagnetic	12 ms				
Bates et al., 1989 (3)	Animals	Electromagnetic	15 ms	Variable Opening shutter	Repetitive		
Liistro et al., 1989 (23)	SB	Pneumatic	6–7 ms	100 ms	Repetitive		
Ohya et al., 1989 (25)	SB	Rotary solenoid and steel gate valve	17 ms	36 ms Variable	Single	<10 ml/s per 10 l/s	
Vooren and van Zomeren, 1989 (34)	SB	Electromagnetic	5 ms	150 ms–4 s	Single		10 ml
Chowienczyk et al., 1991 (8)	SB	Plate driven by servomotor	5 ms	100 ms	Single	<1 ml/s at 2 kPa	
Fletcher et al., 1992 (13)	SB, pediatrics	Manual	<30 ms	Variable Manually operated	Repetitive (6–7/s)		
Hultzsch and Lipowsky, 1992 (19)	MV, pediatrics	Electrical	10 ms	100 ms–1 s	Single		0 ml
Pesenti et al., 1992 (27)	SB	Latex balloon airway valve	10–20 ms	2–3 s	Single, inspiration or expiration		
Smith et al., 1992 (32)	MV, animals	Electromagnetic	5–6 ms	40 ms Variable	Repetitive (10–15 times)		2.4 ml
Freezer et al., 1993 (14)	MV, pediatrics		12 ms	100–500 ms Opening shutter	Repetitive	Pressure tight, 5,280 cmH$_2$O	2 ml
Phagoo et al., 1993 (28)	SB	Elliptical metal plate driven by servomotor (Micro-medical)†	5–6 ms	100 ms	Single		
Van Altena and Gimeno, 1994 (1)	SB	Jaeger pneumoscope‡	5 ms	100 ms	Single		
Carter et al., 1994 (7)	SB, pediatrics	Two rotating disks Rotation at constant frequency	12 ms	100 ms	Repetitive (3/s)		110 ml including PMT
Frey and Kraemer, 1995 (15)	SB, pediatrics	Jaeger bronchoscreen‡	15 ms	100 ms	Single		
Frey et al., 1997 (16)	SB	Rotating blade driven by stepper motor	1 ms	14.5 ms	Repetitive (4 times)		34 ml
Oswald-Mammosser et al., 1997 (26)	SB, pediatrics	Electromagnetic	4 ms	100 ms	Single in inspiration and expiration		75 ml including PMT

MV, mechanical ventilation; SB, spontaneous breathing; CS, computer simulation; PMT, pneumotachograph. Space means data are not available. *Hans Rudolph, Kansas City, MO. †Micro-medical, Ltd., Rochester, UK. ‡Jaeger, Würzburg, Germany.

With a leakage flow of <0.6 ml/s, our valve is placed close to the simulated leakage flow line of 0.5 ml/s (Fig. 3). The relative error in the estimation of Rti would be ~2%. However, in the presence of very low respiratory system compliance, the relative error would rise to ~4%.

These results show that our new valve fulfills both requirements: rapid closing and sufficient tightness in the closed position.

In conclusion, the nonideal technical properties of an occlusion device in combination with reduced respiratory system compliance considerably influence the accuracy in determining Raw and Rti by using the interrupter technique. The existing devices that have been used until now were more or less suitable for determining either Raw or Rti. On the basis of the very limited technical descriptions in the literature, hardly any of the existing devices are suitable for determining both values with sufficient accuracy.

Our newly developed occlusion valve circumvents these problems and can be considered a universal tool for using the interrupter technique, with acceptable errors in the determination of Raw and Rti, in patients with a variety of clinical conditions.

This work may be a further step toward a standardized use of the interrupter technique as proposed by Carter et al. (6).

REFERENCES

1. **Altena, R., Van, and F. Gimeno.** Respiratory resistance measured by flow-interruption in a normal population. *Respiration* 61: 249–254, 1994.
2. **Bates, J. H. T., P. Baconnier, and J. Milic-Emili.** A theoretical analysis of the interrupter technique for measuring respiratory mechanics. *J. Appl. Physiol.* 64: 2204–2214, 1988.
3. **Bates, J. H. T., K. A. Brown, and T. Kochi.** Respiratory mechanics in the normal dog determined by expiratory flow interruption. *J. Appl. Physiol.* 67: 2276–2285, 1989.
4. **Bates, J. H. T., I. W. Hunter, P. D. Sly, S. Okubo, S. Filialtrault, and J. Milic-Emili.** Effect of valve closure time on determination of respiratory resistance by flow interruption. *Med. Biol. Eng. Comput.* 25: 136–140, 1987.
5. **Bates, J. H. T., and J. Milic-Emili.** The flow interruption technique for measuring respiratory resistance. *J. Crit. Care* 6: 227–238, 1991.
6. **Carter, E. R.** It is time to consider standardizing the interrupter technique. *Eur. Respir. J.* 10: 1428–1429, 1997.
7. **Carter, E. R., A. A. Stecenko, B. H. Pollock, and M. J. Jaeger.** Evaluation of the interrupter technique for the use of assessing airway obstruction in children. *Pediatr. Pulmonol.* 17: 211–217, 1994.
8. **Chowienczyk, P. J., C. P. Lawson, S. Lane, R. Johnson, N. Wilson, M. Silverman, and G. M. Cochrane.** A flow interruption device for measurement of airway resistance. *Eur. Respir. J.* 4: 623–628, 1991.
9. **Clarke, J. R., M. J. Jaeger, J. L. Zumrick, R. O'Bryan, and W. H. Spaur.** Respiratory resistance from 1 to 46 ATA measured with the interrupter technique. *J. Appl. Physiol.* 52: 549–555, 1982.
10. **Clements, J. A., J. T. Sharp, R. P. Johnson, and J. O. Elam.** Estimation of pulmonary resistance by repetitive interruption of airflow. *J. Clin. Invest.* 38: 1262–1270, 1959.
11. **Eissa, N. T., V. M. Ranieri, C. Corbeil, M. Chassé, F. M. Robatto, J. Braidy, and J. Milic-Emili.** Analysis of behavior of the respiratory system in ARDS patients: effects of flow, volume and time. *J. Appl. Physiol.* 70: 2719–2729, 1991.
12. **Fichter, J., W. Wierich, and W. Hartung.** Resistance measurement in normal and obstructed excised human lungs by means of the interrupter method. *Respiration* 56: 34–42, 1989.
13. **Fletcher, M. E., C. A. Dezateux, and J. Stocks.** Respiratory compliance in infants—a preliminary evaluation of the multiple interrupter technique. *Pediatr. Pulmonol.* 14: 118–125, 1992.
14. **Freezer, N. J., C. J. Lanteri, and P. D. Sly.** Effect of pulmonary blood flow on measurements of respiratory mechanics using the interrupter technique. *J. Appl. Physiol.* 74: 1083–1088, 1993.
15. **Frey, U., and R. Kraemer.** Interrelationship between postocclusional oscillatory pressure transients and standard lung function in healthy and asthmatic children. *Pediatr. Pulmonol.* 19: 379–388, 1995.
16. **Frey, U., B. Suki, R. Kraemer, and A. C. Jackson.** Human respiratory input impedance between 32 and 800 Hz, measured by interrupter technique and forced oscillations. *J. Appl. Physiol.* 82: 1018–1023, 1997.
17. **Gottfried, S. B., A. Rossi, L. Calverley, L. Zocchi, and J. Milic-Emili.** Interrupter technique for measurement of respiratory mechanics in anesthetized cats. *J. Appl. Physiol.* 56: 681–690, 1984.
18. **Gottfried, S. B., A. Rossi, B. D. Higgs, P. M. A. Calverley, L. Zocchi, C. Bozic, and J. Milic-Emili.** Noninvasive determination of respiratory system mechanics during mechanical ventilation for acute respiratory failure. *Am. Rev. Respir. Dis.* 131: 414–420, 1985.
19. **Hultzsch, W., and G. Lipowsky.** Intrapulmonale Druckmessungen bei Hochfrequenzbeatmung extrem kleiner Frühgeborener. *Monatsschr. Kinderheilkd.* 140: 476–482, 1992.
20. **Jackson, A. C., H. T. Milhorn, and J. R. Norman.** A reevaluation of the interrupter technique for airway resistance measurement. *J. Appl. Physiol.* 36: 264–268, 1974.
21. **Knudson, R. J., J. Mead, and D. E. Knudson.** Contribution of airway collapse to supramaximal expiratory flows. *J. Appl. Physiol.* 36: 653–667, 1974.
22. **Kochi, T., S. Okubo, W. A. Zin, and J. Milic-Emili.** Flow and volume dependence of pulmonary mechanics in anesthetized cats. *J. Appl. Physiol.* 64: 441–450, 1988.
23. **Liistro, G., D. Stanescu, D. Rodenstein, and C. Veriter.** Reassessment of the interruption technique for measuring flow resistance in humans. *J. Appl. Physiol.* 67: 933–937, 1989.
24. **Neergaard, K., Von, and K. Wirz.** Die Messung der Strömungswiderstände in den Atemwegen des Menschen, insbesondere bei Asthma und Emphysem. *Z. Klin. Med.* 105: 51–82, 1927.
25. **Ohya, N., J. Huang, T. Fukunaga, and H. Toga.** Mouth pressure curve on abrupt interruption of airflow during forced expiration. *J. Appl. Physiol.* 66: 509–517, 1989.
26. **Oswald-Mammosser, M., C. Llerena, J. P. Speich, L. Donato, and J. Lonsdorfer.** Measurements of respiratory system resistance by the interrupter technique in healthy and asthmatic children. *Pediatr. Pulmonol.* 24: 78–85, 1997.
27. **Pesenti, A., P. Pelosi, G. Foti, L. D'Andrea, and N. Rossi.** An interrupter technique for measuring respiratory mechanics and the pressure generated by respiratory muscles during partial ventilatory support. *Chest* 102: 918–923, 1992.
28. **Phagoo, S. B., R. A. Watson, N. B. Pride, and M. Silverman.** Accuracy and sensitivity of the interrupter technique for measuring the response to bronchial challenge in normal subjects. *Eur. Respir. J.* 6: 996–1003, 1993.
29. **Romero, P. V., J. Sato, F. Shardonofsky, and J. H. T. Bates.** High-frequency characteristics of respiratory mechanics determined by flow interruption. *J. Appl. Physiol.* 69: 1682–1688, 1990.

30. **Schlesinger, R. B., A. V. Zeccardi, and J. Monahan.** An interrupter method for measurement of airway resistance in small animals. *J. Appl. Physiol.* 48: 1092–1095, 1980.

31. **Sly, P. D., and J. H. T. Bates.** Computer analysis of physical factors affecting the use of the interrupter technique in infants. *Pediatr. Pulmonol.* 4: 219–224, 1988.

32. **Smith, P. G., A. Falahat, and W. A. Carlo.** Measurement of interrupter resistance in rabbits exposed to methacholine aerosols. *J. Appl. Physiol.* 72: 2454–2457, 1992.

33. **Tantucci, C., C. Corbeil, M. Chassé, F. M. Robatto, S. Nava, J. Braidy, N. Matar, and J. Milic-Emili.** Flow and volume dependence of respiratory system flow resistance in patients with adult respiratory distress syndrome. *Am. Rev. Respir. Dis.* 145: 355–360, 1992.

34. **Vooren, P. H., and B. C. Van Zomeren.** Reference values of total respiratory resistance, determined with the "opening" interruption technique. *Eur. Respir. J.* 2: 966–971, 1989.