Passive expiration as a test of lung function

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Ashutosh, K, and Keighley, J F (1978). *Thorax, 33, 740–746. Passive expiration as a test of lung function.* Twenty-five normal subjects, 14 non-smokers and 11 smokers, passively expired into a spirometer after a maximal active inspiration, and after a passive inflation of the chest by a pressure cycled intermittent positive-pressure breathing (IPPB) machine. Acceptable passive expirations could be performed by all subjects after a passive inspiration but by only 12 after an active inspiration. Expired volume was found to change exponentially with time ($t > 0.98$), and the time constant of passive expiration ($T_p$) was obtained. There was no significant difference between the smokers and non-smokers in age, sex, forced vital capacity, FEV$_1$, FEV$_1$/FVC $\%$, maximum mid-expiratory flow rate, maximum expiratory flow at 50 and 25 $\%$ of the vital capacity, or the magnitude of the fall in the dynamic compliance with increasing frequency of breathing ($C_{dyn/f}$). $T_p$ in smokers ($1.06 s \pm 0.47 SD$) was significantly longer than in the non-smokers ($0.65 s \pm 0.25 SD$ $p < 0.02$). $T_p$ had a significant correlation with $C_{dyn/f} (T_p = 0.6 + 161.81 C_{dyn/f} \pm 0.38 SE, r=0.49, p < 0.02)$. We conclude that satisfactory passive expiratory spirometers can be easily obtained after a mechanically assisted passive inspiration. $T_p$ thus obtained is determined by the intrinsic properties of the respiratory system (lung plus thorax), and is significantly prolonged in smokers compared with non-smokers when other studies of pulmonary function including frequency dependence of compliance are unchanged.

Passive expiration, defined as the expiration when the lungs and thorax are allowed to return to functional residual capacity (FRC) by their own elastic recoil and no active muscular effort is applied, has been shown to be determined solely by the intrinsic properties of the respiratory system (DuBois and Ross, 1951). Although found to be a useful method of measuring total respiratory (lung + chest wall) compliance and resistance (Brody et al, 1960), it has not been widely utilised because of the difficulty in obtaining a passive expiration in unanaesthetised, untrained individuals (Agostoni and Mead, 1964).

We found that passive expirations were more consistently and easily obtained if preceded by a passive inflation of the chest by an intermittent positive-pressure breathing machine (IPPB). The present study was undertaken to determine the practicality of obtaining passive spirometers in untrained individuals by the above method. We also tested the ability of the passive spirometer to detect early abnormalities in lung function compared with other pulmonary function tests including frequency dependence of compliance and flow-volume curves.

Methods

Twenty-five normal healthy volunteers were selected for the study. Four subjects, two of whom were smokers, had a mild morning non-productive cough. None had any known pulmonary illness at the time of the study. All had had a normal expiratory spirogram and a normal chest radiograph within one year of the study. Fourteen were non-smokers (men 8, women 6) and eleven were smokers (men 8, women 3). The difference in the age and sex composition between the smokers and non-smokers was not significant (see table).

Subjects reported to the laboratory in a fasting state. Simultaneous forced expiratory spirometers and expiratory flow volume curves were obtained with a low resistance wedge spirometer (Med Science Electronics—Wedge Spirometer Model 370), with electronic output fed into two X-Y plotters (HP X-Y Recorders, Model 7046, Hewlett Packard Co, Waltham, Mass).

Next, a 10 cm long latex oesophageal balloon attached to the end of a 100 cm polyethylene tube having multiple holes in its distal 9 cm was passed into the distal one-third of the oesophagus through
the mouth or nose. The proximal end of the polyethylene tube was connected to a pressure transducer, and the pressure was recorded on a multichannel electronic recorder (HP Sanborn Recorder, Hewlett Packard). The final position of the balloon was that which showed the least amount of cardiac artifact and maximum deflections in the pressure signals during respiration.

With the oesophageal balloon in position, the passive spirogram was recorded by two different methods: (1) the subject inspired maximally and at end-inspiration suddenly relaxed his chest muscles, glottis, and pharynx, and let his chest deflate passively; (2) a Bennett PR2 IPPB machine (Puritan Bennett Co, Berkeley, California) was set at a pressure of 20 cm of water and was connected to the subject through a mouthpiece. The subject sealed his lips tightly around the mouthpiece, and the system was checked for any leaks. The subject initiated an inspiration and then immediately relaxed his respiratory muscles, pharynx, and glottis as completely as possible, and allowed the machine to force air into his chest at the preset pressure. At the end of the positive-pressure inspiration he continued to relax and allowed his chest to deflate passively without using any effort. Thus in this method both inspiration and expiration were passive. During these manoeuvres simultaneous intraoesophageal pressure recordings, volume-time spiroms, and flow volume curves were obtained by connecting the expiratory line of the IPPB machine to a wedge spirometer-X-Y recorder system as in the case of the active spirogram.

Two criteria were used to determine whether the expirations were completely passive. One was a rapid (within 0.4 seconds from the onset of the expiration) rise or fall of the intraoesophageal pressure to the pressure previously recorded at the FRC, usually −1 or −2 cm of water. The second criterion was a linear expiratory limb of the flow volume curve of the spirogram. We analysed only those expirations that satisfied both the criteria, except in the one instance when oesophageal pressure was not recorded (figs 1 and 2).

To measure the frequency dependence of compliance, the oesophageal pressure and the air flow at the mouth were recorded on a multichannel (Sanborn Div, Hewlett-Packard Co, Waltham, Mass) recorder, and the volume signal was obtained by integrating the flow signal from a pneumotachometer. The subjects, who were asked to breathe at different frequencies (f) with tidal volume in the range of 0.6 to 1.2 l, inspired to the maximum inspiratory capacity and then expired to FRC between breathing at different frequencies. An effort was made to keep the end expiratory lung volume constant by instructing the patient to control his breathing so as to keep the end-expiratory point in the volume tracing at the same level. Neither the volume nor the end-expiratory point, however, could be kept completely identical at all the breathing frequencies. Compliance at different frequencies was measured by dividing the change in volume by the change in pressure between two zero flow points in each respiratory cycle. Later, we used a computer channel (HP 8816A—Respiratory Analyzer, Hewlett Packard) that directly recorded dynamic lung compliance at each breathing frequency. Readings of compliance obtained by the computer were always randomly checked, at least once in each breathing frequency, against the value obtained by dividing volume by pressure change as displayed in the volume and pressure recording. Frequency dependence of compliance could not be measured in one subject because of her inability to swallow the oesophageal balloon.

Age, sex, and smoking history in pack years were recorded. Forced vital capacity (FVC), FEV1, maximal mid-expiratory flow rate (FEF25−75%), and FEV1/FVC% were calculated from the active spirogram. Maximum flow at 50% of vital capacity

### Table: Results (mean ± 1SD) of selected tests in smokers and non-smokers

<table>
<thead>
<tr>
<th></th>
<th>Smokers</th>
<th>Non-smokers</th>
<th>Student's two-tailed t test</th>
</tr>
</thead>
<tbody>
<tr>
<td>No of subjects</td>
<td>11</td>
<td>14</td>
<td></td>
</tr>
<tr>
<td>Age (years)</td>
<td>30.27±8.84</td>
<td>25.64±4.72</td>
<td>NS</td>
</tr>
<tr>
<td>FEV1 (l)</td>
<td>3.43±0.80</td>
<td>3.58±0.57</td>
<td>NS</td>
</tr>
<tr>
<td>FVC (l)</td>
<td>4.31±1.07</td>
<td>4.16±0.68</td>
<td>NS</td>
</tr>
<tr>
<td>FEV1/FVC (%)</td>
<td>80.36±6.84</td>
<td>84.93±7.05</td>
<td>NS</td>
</tr>
<tr>
<td>Vmax40 (l/sec)</td>
<td>3.59±1.26</td>
<td>4.21±1.09</td>
<td>NS</td>
</tr>
<tr>
<td>Vmax50 (l/sec)</td>
<td>3.78±1.31</td>
<td>4.45±1.28</td>
<td>NS</td>
</tr>
<tr>
<td>PEV1 (l)</td>
<td>1.10±0.24</td>
<td>1.19±0.27</td>
<td>NS</td>
</tr>
<tr>
<td>Tp (s)</td>
<td>1.06±0.47</td>
<td>0.65±0.25</td>
<td>NS</td>
</tr>
<tr>
<td>Ta (s)</td>
<td>0.87±0.54</td>
<td>0.54±0.16</td>
<td>NS</td>
</tr>
<tr>
<td>Co (/l/cm water)</td>
<td>0.22±0.10</td>
<td>0.23±0.08</td>
<td>NS</td>
</tr>
<tr>
<td>Cdyn/F (/cm H2O/sec−1)</td>
<td>1.7 ±1.3</td>
<td>1.2 ±1.3</td>
<td>NS</td>
</tr>
</tbody>
</table>

NS = Not significant.

For abbreviations, see text.
(V max30) and at 25\% of vital capacity (V max25), and peak flow (V max) were obtained from the maximum expiratory flow volume curve. Passive expired volume in one second (PEV1) was obtained from the passive expiratory spirogram.

The active and passive spiromgrams were replotted on a semilogarithmic paper, the time on the X axis, and the unexpired volume as a percentage of the total expired volume

\[
\frac{\text{total expired volume} - \text{expired volume to time } t\%}{\text{total expired volume}}
\]

on the Y axis at 0.2 second intervals.

The change in expired volume was exponentially related to time (t \geq 0.98). The relationship between the expired volume and time is described by the equation \(V_t = V_0 e^{-t/T}\) where \(V_t = \text{unexpired volume at a given time } t, V_0 \text{ the initial volume and } T = \text{the time constant of the total respiratory system (lung + thorax). At the instant when } t = T, V_T = V_0 e^{-1} = V_0 0.37 (V_T = \text{unexpired volume up to the instant when } t = T).\) Therefore, the time at which the unexpired volume falls to 0.37 of the original value equals the time constant. Time constants of active expiration (T_A) and passive expiration after a maximal inspiration (T_{P_{max}}) and after a passive inspiration by IPPB device (T_p) were thus calculated.

Dynamic compliance on Y axis was plotted against different breathing frequencies on X axis. The regression line extrapolated to zero yielded the estimated compliance of the lung at zero f (CO). The slope of the regression line (Cdyn/f) was taken as the measure of the frequency dependence of compliance.

**Results**

All subjects produced many successive, satisfactory passive expired after a few trials when the inspiration was passive and accomplished by IPPB. On the other hand, it was difficult for them to relax adequately for an acceptable passive expiration when the inspiration was active. Thirteen subjects (52\%) could not produce a single satisfactory passive expiration after active inspiration, and many subjects could learn to relax sufficiently only after many trials. Passive expired after passive inflation of the chest by IPPB always met the criteria for adequate passive expired if the multiple spiromgrams thus obtained were fully superimposable.

The intraoesophageal pressure at the end of a passive inflation of the chest ranged between +4 and +10 cm H2O in most subjects. In one subject the end-inflation pressure rose to 22 cm H2O, probably from the subject resisting the positive pressure.

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**Fig 1** Examples of oesophageal pressure recordings in acceptable passive expired. Time is recorded along X axis and pressure (cm of H2O) is displayed on Y axis. I and E denote inspiration and expiration respectively. (a) and (b) were obtained by method of passive inspiration and (c) after active inspiration. Note rapid change of oesophageal pressure to FRC.

**Fig 2** Flow volume curves of passive expiration: (a) is after an active inspiration while (b) is after successive passive inspirations.
inflation by contracting his thoracic muscles (fig 1a). Nevertheless, the following expiration still met the criteria for an adequate passive expiration.

Subjects were divided into two groups—smokers and non-smokers. The table summarises the results of selected tests in the two groups.

Time constant of passive expiration \( (T_p) \) was significantly prolonged in smokers compared to the non-smokers \( (p < 0.05, \text{table}) \). Figure 3(a) shows that 73\% of the smokers and only 21\% of the non-smokers had a \( T_p \) of 0.7 or more \( (p < 0.05) \). Mean \( \text{Cdyn}/f \) was also lower in the smokers than in the non-smokers, but the difference was not statistically significant (see table).

No other difference between the two groups was observed.

The relation between dynamic compliance and respiratory frequency was linear \( (\text{Mean } r = 0.86) \) except in one subject in whom the dynamic compliance at highest breathing frequency was unchanged from that at \( f < 10 \).

\( T_p \) and \( \text{Cdyn}/f \) showed a significant correlation \( (T_p = 0.60 + 161 \cdot 81 \text{Cdyn}/f \pm 0.38 \text{ SE}, r = 0.49, p < 0.02) \) (fig 4).

Eleven subjects had \( \text{Cdyn}/f \) of equal to or less than 0.001, and 13 had values larger than 0.001. Although a higher proportion (54\%) of subjects with \( \text{Cdyn}/f \) > 0.001 smoked than of those with \( \text{Cdyn}/f \leq 0.001 \) (27\%), the difference was not significant. \( T_p \) was significantly longer in the subjects with \( \text{Cdyn}/f \) > 0.001 \( (1.06 \pm 0.46 \text{ SD}) \) than in the group with \( \text{Cdyn}/f \leq 0.001 \) \( (0.59 \pm 0.17 \text{ SD}, p < 0.01) \). Figure 3(b) shows that 54\% of the subjects with \( \text{Cdyn}/f \) of more than 0.001 and none of the subjects with \( \text{Cdyn}/f \) of 0.001 or less had a \( T_p \) greater than one second \( (p < 0.025) \). Similarly, of the 17 subjects with \( T_p > 1 \text{ s} \) 65\% had \( \text{Cdyn}/f \) of 0.001 or less. Of the seven subjects with \( T_p > 1 \text{ s} \) none had a \( \text{Cdyn}/f \) of 0.001 or less.

No significant difference was noted between \( T_p \) and \( T_{p \text{max}} \). Although mean time constant of active

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**Passive expiration as a test of lung function**

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**Fig 3(a)** Time constant of passive expiration \( (T_p) \) in smokers and non-smokers; (b) \( T_p \) in subjects with \( \text{Cdyn}/f \) of more, and less than 0.001; and (c) Time constant of active expiration \( (T_a) \) in smokers and non-smokers. Mean and ±1 SD are shown by solid and broken lines respectively.

**Fig 4** Correlation between \( T_p \) and \( \text{Cdyn}/f \). Regression line and ±1 SE are shown. Closed circles denote smokers and open circles non-smokers.
expiration \( (T_A) \) was shorter in both smokers \( (0.87 \text{ s} \pm 0.59 \text{ SD}) \) and non-smokers \( (0.54 \text{ s} \pm 0.16 \text{ SD}) \) compared to \( T_B \), the difference between \( T_A \) and \( T_B \) was not significant (fig 3(c)).

**Discussion**

Free collapse of the lung, that is, when the lungs deflate solely by the force of their elastic recoil, has been shown to reflect the intrinsic properties of the respiratory system. A passive expiration, when no muscular force is applied, can be used as a model reflecting free collapse of the lung plus thorax (Ebert and Haddy, 1954; Nissel and DuBois, 1954). Dixon and Brodie (1903) were first to use free collapse of the lung in animal experiments to show changes in lung mechanics. Since then passive expiration has been studied in isolated lungs (Nissel, 1948), anaesthetised human subjects (Newman et al, 1959), and trained volunteers (Otis et al, 1950), and has been recommended as a means for measuring total compliance and resistance of the lung plus thorax in man (Comroe et al, 1954).

It can be shown by mathematical analysis that during a passive expiration the change in volume in relation to time is exponential, and the ratio between the change in volume and the change in time at any given instant must be constant and equal to the product of the total respiratory (lungs + chest wall) resistance and compliance, that is, the time constant. Furthermore, the volume time course of a passive expiration is determined solely by the volume elastic and flow resistive characteristics of the respiratory system, and it has a fairly constant half-time predictable on the basis of the above properties (Pierce, 1959; McIlroy et al, 1963).

Despite the above works confirming the theoretical and experimental validity of the tests based on passive expiration, we are not aware of any study using passive spirometry as a test of pulmonary function in man and evaluating its relative sensitivity in comparison with other tests of the lung function, although the clinical usefulness of passive spirometry in patients with acute respiratory failure was suggested by Ashutosh et al (1975). One reason for this may be the difficulty in obtaining a reliable passive expiration in unanaesthetised subjects. Brody (1954) has shown that a voluntary passive expiration is virtually impossible because of persisting inspiratory tone in the respiratory muscles during a relaxed expiration. This tone of the inspiratory muscles becomes negligible only towards the end of the relaxed expiration if the previous inspiration is an active process. Probably when inspiration is a passive process there will be much less or even absent inspiratory tone during the following ‘relaxed’ expiration. This would explain the ease with which our subjects could perform satisfactory passive expirations after IPPB inspirations.

The criteria defining an acceptable passive expiration have been previously established by other workers. Pierce (1959) considered a rapid fall of oesophageal pressure to the baseline as denoting a passive expiration. McIlroy et al (1963) showed that a linear descending slope of the expiratory flow volume curve would correspond to a completely passive expiration. Both of the above criteria were met in all the acceptable passive expirations in our study.

Because asymptomatic smokers with normal routine pulmonary function tests are known to have more abnormalities compared with normal non-smokers when more sensitive tests are applied, the ability of a test to detect deviations from normal in smokers is widely considered to reflect its ability to detect early changes in pulmonary mechanics, presumably due to small airways obstruction (Dosman et al, 1975). \( \text{FEF}_{25-75\%} \), \( \text{Vmax}_{50} \) and \( \text{Vmax}_{25} \) (Tockman et al, 1967; Marcq and Minette, 1976), and in this respect agree with the results obtained in our subjects.

Frequency dependence of compliance is considered to be the most definitive test for small airways disease and has been the standard test against which all others have been compared (Woolcock et al, 1969). Guyatt et al (1975) pointed out that although all workers agree that dynamic compliance \( (C_{dyn}) \) falls at higher frequency of breathing \( (f) \) in the presence of obstructive lung disease, only a few have quantified frequency dependence. Furthermore, the method of expressing frequency dependence of \( C_{dyn} \) has varied with different observers. Mills et al (1963) used the slope of the regression line of \( C_{dyn} \) against \( f \), Woolcock et al (1969) measured the value of \( C_{dyn} \) at the fastest \( f \) attained in a subject as a percentage of the static compliance of lung \( (C_{st}) \), McFadden et al (1974) have used the same value but at \( f \) of 60/min, while Ingram and O’Cain (1971) divided the values of \( C_{dyn} \) at all the breathing frequencies by \( C_{st} \), for quantifying the magnitude of frequency dependence of \( C_{dyn} \). Comparison of \( C_{dyn} \) with \( C_{st} \) may produce inaccuracies because of the difference in the tissue and surface hysteresis of the lung between static and dynamic conditions (Saibene and Mead, 1969).
Passive expiration as a test of lung function

In view of the uncertainties discussed above, spirometry, MEFV curves, and frequency dependence of compliance fail to do so.

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