Eur Respir J 2002; 19: 530–533
DOI: 10.1183/09031936.02.00227102

European Respiratory Journal
ISSN 0903-1936

Recording flow in the first second of a maximal forced expiratory manoeuvre: influence of frequency content

M.R. Miller*, J. Lloyd#, P. Bright¶

Recording flow in the first second of a maximal forced expiratory manoeuvre: influence of frequency content. M.R. Miller, J. Lloyd, P. Bright. ©ERS Journals Ltd 2002.
ABSTRACT: The frequency content of the first second of the maximum forced expiratory manoeuvre (MFEM) was measured to determine if the currently accepted frequency limit of 20 Hz for MFEM is adequate for recording peak expiratory flow (PEF).

The frequency response of a Fleisch pneumotachograph (PT) was measured and used to record MFEM from 24 patients attending a lung-function laboratory and 26 normal volunteers. The first 1.024 s of the signal recorded at 1,000 Hz for that blow with maximum PEF, underwent fast Fourier transformation using a triangular window function, applied after 0.75 s to reduce flow linearly to zero. All the frequencies above a set limit were removed, followed by inverse transformation to reconstitute the blow. The limits for this frequency cut-off were progressively varied from 100 Hz down to 15 Hz, with the resulting PEF being compared with the PEF from the reconstituted blow with no frequency reduction.

The average±SD age for the group was 47±18 yrs and the average PEF was 450±187 L·min^{-1}, which, when expressed as a standardized residual, was 0.1±2.1, with a range from -4.5–3.9 indicating a good spread around normal values. Average rise time to PEF was 83±38 ms and dwell time >90% PEF was 45±25 ms. Cut-off >20 Hz reduced the mean PEF of the group by 8.5 L·min^{-1} (95% confidence limit 5.5–11.4 L·min^{-1}), whereas cut-off >30 Hz reduced mean PEF by 4.4 L·min^{-1} (2.6–6.2). In the present study subjects, 30 Hz was on the 95th percentile of frequencies for defining the upper limit for 98% of the power spectrum for the first second of the blow.

It has been shown that there are frequencies >20 Hz that contribute to peak expiratory flow enough to influence readings made using conventional hand-held peak expiratory flow meters, such as the mini-Wright. Devices used for recording flow from a maximum forced expiratory manoeuvre should therefore have an adequate frequency response of up to 30 Hz.
Eur Respir J 2002; 19: 530–533.

*Dept of Medicine, University of Birmingham, Selly Oak Hospital, Birmingham, #Dept of Medicine, Good Hope NHS Trust, Sutton Coldfield, West Midlands and ¶Dept of Medicine, Solihull NHS Trust, Solihull, West Midlands.

Correspondence: M.R. Miller
Dept of Medicine
University of Birmingham
Selly Oak Hospital
Birmingham B29 6JD
UK
Fax: 44 1216278292
E-mail: m.r.miller@bham.ac.uk

Keywords: Fast Fourier analysis
lung function
peak expiratory flow
pneumotachography
spirometry

Received: March 18 2001
Accepted after revision August 22 2001

The frequency content of the maximal forced expiratory manoeuvre (MFEM) has been previously defined, with the finding of a significant amplitude content in frequencies up to 20 Hz for maximal breathing-capacity manoeuvres [1] and up to 4 Hz for tidal breathing and vital capacity manoeuvres recorded with a bell spirometer. Other studies found that <5% of the total frequency content was >10 Hz in males performing forced expiratory flow manoeuvres [2]. Therefore, 20 Hz has been accepted as the limit for adequate recording devices when measuring the whole MFEM [3]. One of the lung-function indices that might be most affected by alterations in frequency content is peak expiratory flow (PEF). It is possible that the frequency content at the beginning of an MFEM that contributes to PEF, may be higher than that suggested from studying the MFEM overall. This aspect of lung function may not be recorded with adequate fidelity if this fact is ignored.

Therefore, this study was undertaken to determine the frequency content of the first second of the MFEM, so that endeavours to test the frequency-response characteristics of PEF recording devices can be properly matched to the signal requirements.

Method

The study used an optimized pneumotachograph (PT), which comprised a Vitalograph PT head and upstream geometry, with the mesh screen inserted upstream to the PT head to improve its linearity [4]. The PT head was unheated and thermally stabilized by placing it on a fan in between blows [5]. A Sensym differential low-pressure transducer (type SCXL004DN; Farnell Electronics Components Ltd, Leeds, UK) was used, with the signal amplified and low-pass filtered with a Butterworth 4 pole filter set at 200 Hz. Based on the assumption that the whole PT assembly, including the transducer, would behave as a second-order system, its frequency response was tested in the following manner using a step test [6]. A flow was delivered from an explosive decompression system [7], which was then suddenly terminated using a computer-controlled fast-response direct drive valve (type D633–313A; Moog Inc., NY, USA), the closure time from fully open was <12 ms. The transducer's output was sampled at 1,000 Hz with a 12-bit analogue-to-digital converter and stored. The transducer's frequency response was also tested on its own by a step test. A glass syringe was attached to one port of the transducer and the plunger was withdrawn, giving a strong negative pressure that was suddenly released to atmospheric pressure. The output was sampled at 1,000 Hz and stored. Measurements were made of the amplitude of the first two oscillations of these recordings (amp1 and amp2) as they settled to zero, and the periodicity (Td s) of the oscillations was determined. The natural resonant frequency (Fn) and damping coefficient (D) were calculated from these measurements using the following formulae [6], which assumes a second-order system response:

$$\mathrm{k} = \log_e(\mathrm{amp1}/\mathrm{amp2}) \quad (1)$$

$$\mathrm{D} = \mathrm{k}/\sqrt{(4\times\pi^2+\mathrm{k}^2)} \quad (2)$$

$$\mathrm{Fn} = 1/\left(\mathrm{Td}\times\sqrt{(1-\mathrm{D}^2)}\right) \quad (3)$$

The PT was calibrated before each recording session using a 3-L syringe and the method of VARENE *et al.* [8]. The syringe was emptied through the PT at least twice with flows in the range of 1–3 $L \cdot s^{-1}$, then twice at 4–6 $L \cdot s^{-1}$, followed by twice at 7–12 $L \cdot s^{-1}$. These data were used to derive an average calibration factor, thus avoiding any bias toward a particular range of flow. The linearity of the system was measured by delivering constant known flows between 0.5–12 $L \cdot s^{-1}$ using a pump system [9].

This PT system was used to record MFEM from 50 volunteer subjects. Twenty-four of these subjects were patients (11 males) attending a lung-function laboratory for routine testing and 26 (13 males) were volunteers from hospital staff. Three MFEM that met the acceptance criteria of the American Thoracic Society (ATS) [10] were recorded for each subject. Flow was sampled at 1-ms intervals and stored on computer disk. From the three blows, maximal PEF, forced vital capacity (FVC) and forced expiratory volume in one second (FEV_1) were calculated and related to predicted values using the European Community for Steel and Coal (ECSC) equations and the method of standardized residuals [3]. The blow with the largest PEF had a 10–90% rise time to PEF and a dwell time for flow >90% of peak calculated. The data from the first 1.024 s of this blow underwent fast Fourier transform (FFT) analysis using Origin version 6.1 (OriginLab. Corp., MA, USA) with a triangular window function applied, so that from 0.75 s flow was linearly reduced to zero at 1.024 s. This was necessary in order to avoid spurious high-frequency content, due to the sudden transition to zero flow by truncation at 1.024 s. The derived FFT coefficients then underwent inverse transformation to regenerate the blow, with no alteration to the coefficients. PEF was derived from this reconstituted blow and called PEF_R. This process of inverse transformation to reconstitute the blow was sequentially repeated with prior removal of all the frequency components at >100, 90, 80, 70 , 60, 50, 40, 30, 20, and 15 Hz. The PEF for these regenerated blows was calculated and stored as PEF_{100}–PEF_{15}, respectively. Comparison between PEF_R and each of the frequency-reduced PEF was performed by paired t-tests.

Any possible effect upon the recorded PEF from the frequency response of the PT system was checked by dividing the amplitude coefficients of the FFT from the flow/time data by the transfer function of the PT gain for each frequency and then performing an inverse FFT. This was only an approximation for the correction, since no account of phase changes could be taken.

Results

The PT was linear across the range 0.5–12 $L \cdot s^{-1}$, with residual SD from the regression line of 0.04 $L \cdot s^{-1}$. Three step tests were performed for the whole PT system, with the Td measured at 5.6, 5.7 and 7.7 ms (mean 6.3 ms), and amplitude ratios of 1.93, 1.82, 1.81 (mean 1.85). The average natural resonant frequency was 159 Hz, with a damping coefficient of 0.1 and a 3 decibel (dB) point of 87 Hz. Figure 1 shows the signal settling after sudden termination of the flow for one of these tests and figure 2 shows the plot-of-gain *versus* frequency for the PT system as a whole. When the transducer and its tubing were tested on their own, the resonant frequency was 333 Hz with a damping factor of 0.2.

The FVC, PEF, and FEV_1 data for the normal subjects and the patients, are shown in table 1. PEF_R, which was derived from FFT and inverse FFT of the recorded data, was identical to the raw PEF in $L \cdot s^{-1}$ to three decimal places. Using the approximate correction of the flow/time data for the known gain of the PT system reduced the PEF by an average of only 0.32 $L \cdot min^{-1}$ (0.005 $L \cdot s^{-1}$), with a standard deviation of 0.36 $L \cdot min^{-1}$ and 95% confidence limit (CL) of

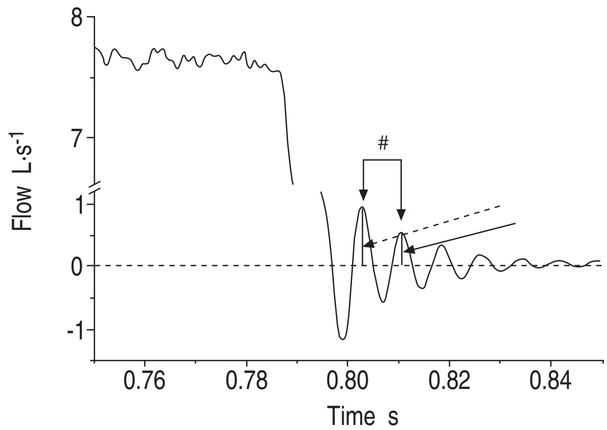

Fig. 1.– Plot of the transducer signal as it settles after cessation of flow through the pneumotachograph assembly. #: periodicity; broken arrow: amplitude 1; solid arrow: amplitude 2.

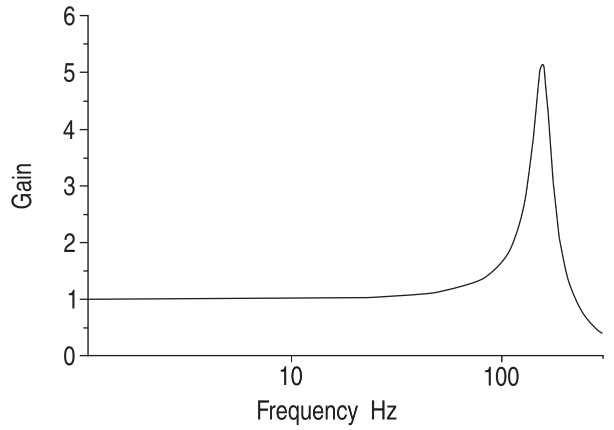

Fig. 2. – Plot of the gain *versus* frequency for the whole pneumotachograph assembly. Resonant frequency: 159 Hz; damping coefficient: 0.1; 3 decibels (dB): 87 Hz is the frequency at which the signal gain is 3 dB.

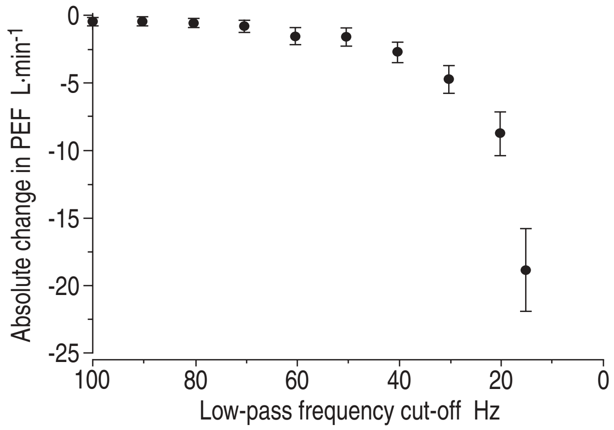

Fig. 3. – Plot of absolute change in peak expiratory flow (PEF) with different cut-off values for fast Fourier transform low-pass filtering of the 50 recorded blows.

Table 1. – Data recorded for the 50 subjects in the study

	Mean±SD	Minimum	Maximum
Age yrs	47.18±18.3	17	83
Height m	1.66±0.09	1.49	1.87
PEF $L \cdot s^{-1}$	7.52±3.10	2.07	14.23
PEF $L \cdot min^{-1}$	451±186	124	854
PEFSR	0.05±2.10	-4.49	3.93
FEV_1 L	2.93±1.34	0.62	5.87
FEV_1SR	-0.22±1.88	-5.41	2.75
FVC L	3.98±1.62	0.75	6.85
FVCSR	0.60±1.91	-5.52	3.98
RT ms	82.6±38.0	30	218
DT ms	44.5±25.4	5	141

PEF: peak expiratory flow; FEV_1: forced expiratory volume in one second; FVC: forced vital capacity; RT: 10–90% rise time for PEF; DT: dwell time of PEF >90% of PEF; PEFSR, FEV_1SR and FVCSR: deviation of PEF, FEV_1 and FVC from their predicted values when expressed as standardized residuals [3].

0.22–0.43 $L \cdot min^{-1}$. Figure 3 shows the reduction in average PEF for the group, with standard error bars for progressive frequency content reduction. When all frequencies >60 Hz were removed, the first statistically significant reduction in mean PEF (95% CL), of 1.2 $L \cdot min^{-1}$ (0.2–2.1 $L \cdot min^{-1}$; $p<0.01$), was seen for the group. However, the first clinically relevant reduction in PEF was with reduction of the frequency content >20 Hz, which caused a mean reduction in PEF of 8.5 $L \cdot min^{-1}$ (5.5–11.4 $L \cdot min^{-1}$; $p<0.00001$). Loss of frequencies >30 Hz caused PEF to drop by a mean of 4.4 $L \cdot min^{-1}$ (2.6–6.2 $L \cdot min^{-1}$; $p<0.00001$). The absolute drop in PEF at any one of the reductions in frequency content was not significantly correlated with original PEF for the subjects tested.

Figure 4 shows a plot of selected percentiles for the frequency that defines various percentages of the power spectrum for the blows from the 50 subjects. The 95th percentile for the subjects was 20 Hz for 97% of the power spectrum and ~30 Hz for 98% of the power spectrum.

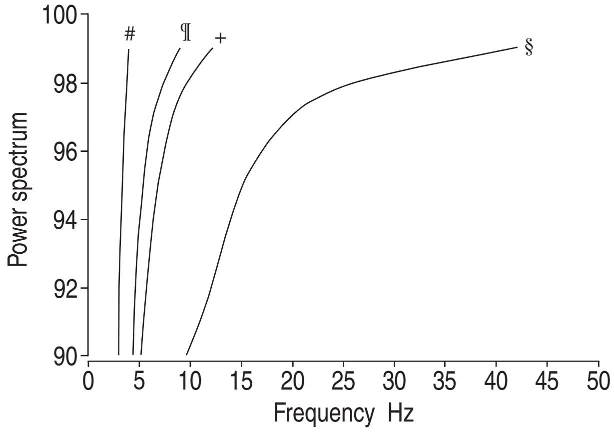

Fig. 4. – Plot of selected percentiles for the 50 subjects for the upper-frequency limit that defines various percentages of the power spectrum for the first second of the blows. #: 2.5th percentile; ¶: 50th percentile: +: 75th percentile: §: 95th percentile.

Discussion

It has been demonstrated that the PEF recorded from the present range of patients and normal subjects contains a significant contribution from frequencies above the accepted level of 20 Hz. The mean reduction in PEF with a cut-off >20 Hz was greater than the limits of accuracy for the usual PEF meters that patients use, and so would alter the reading obtained. With a cut-off >30 Hz, the average difference of 4.4 $L \cdot min^{-1}$ is at the limit of accuracy for readings from these meters. This indicates that, in order to record PEF faithfully, the recording device must be able to handle frequencies up to 30 Hz. There was no correlation between reduction in PEF and the size of the original PEF. Thus, the clinical importance of this reduction may differ between subjects, but the accuracy of the measuring devices will be affected.

To be certain that these conclusions are correct, the recording device for the original signals must be more than adequate for the purpose. The linearity of the PT has been previously verified using an adequate pump system [9], and its calibration has been performed by an accepted method that minimizes any bias across its range. The frequency response has been tested with a step test, which is one of the best ways to determine its characteristics. The method described is suitable for any device that has a continuous analogue output. An explosive decompression device fitted with a fast-response solenoid was used, but other flow-generating devices could be used. An "off" step test was used, as suddenly ceasing a flow has fewer potential artefacts when compared with suddenly generating a constant flow for an "on" step response.

The authors used a system that was under-damped, but with a natural resonant frequency well in excess of the frequency content of interest. An alternative would have been to have a system that had a lower resonant frequency but was critically damped. The authors believe that the PT system was more than adequate for the purpose of recording MFEM from humans. This has been verified by demonstrating the insignificant change in PEF when making an approximate correction for the PT′s frequency-response characteristics. When using this PT system for subjects in a clinical setting, a low-pass analogue filter set at 50 Hz was employed. For this study, a much higher limit was used with a 200 Hz low-pass filter to ensure that none of the frequencies of interest could be affected.

The second requirement for this study to be representative was that the subject population adequately reflected the relevant client group. A sample of 50 subjects was examined. One-half of the sample were without symptoms or known disease and thus regarded as normal, and the other half were symptomatic subjects with airflow limitation. The results in table 1 indicate that these subjects ranged from -4.5–3.9 SD from predicted in PEF, with an average that was essentially as predicted. Whilst a larger sample might have shown tighter confidence limits in these findings, it is believed that the subjects were sufficiently representative for the findings to be accepted. The statistical significance of a change in PEF was at a cut-off frequency of 60 Hz, but it was at 30 Hz that the average change exceeded 5 $L \cdot min^{-1}$, which would be the amount necessary to influence a reading from an analogue meter, such as the mini-Wright.

The present finding is important for verification of whether flow meters have an adequate frequency response to record peak expiratory flow and other flow phenomena. For meters such as the mini-Wright, which only have a single peak-flow output, test signals that span the range of input signals of interest are needed. The characteristics of these signals are probably best described by their rise and dwell time [11]. Such signals, with different frequency content but identical flow, can be delivered to the meters, which should give the same reading irrespective of the frequency content of the input signals. Such tests can be verified by using a pneumotachograph to record the flow. This study presents evidence that the requirements for frequency response testing with regard to measuring peak expiratory flow should include frequencies up to 30 Hz.

Acknowledgements. The authors would like to thank P. Atkins, Dept of Electronic and Electrical Engineering, University of Birmingham, for advice about this project.

References

1. McCall CB, Hyatt RE, Noble FW, Fry DL. Harmonic content of certain respiratory flow phenomena of normal individuals. *J Appl Physiol* 1957; 10: 215–218.
2. Peslin R, Jardin P, Bohadana A, Hannhart B. Contenu harmonique du signal de débit pendant l′expiration forcée chez l′homme normal. *Bull Eur Physiopathol Respir* 1982; 18: 491–500.
3. European Respiratory Society Official Statement. Standardized lung function testing. *Eur Respir J* 1993; 6: Suppl. 16, 5–40.
4. Miller MR, Pincock AC. Linearity and temperature control of the Fleisch pneumotachograph. *J Appl Physiol* 1986; 60: 710–715.
5. Miller MR, Sigsgaard T. Prevention of thermal and condensation errors in pneumotachographic recordings of the maximal forced expiratory manoeuvre. *Eur Respir J* 1994; 7: 198–201.
6. Andersen HR, Bergsten O. Blood Pressure: Measurements and Methods. 2nd Edition. Albertslund, Denmark, S & W Medico Teknik A/S, 1987.
7. Pedersen OF, Miller MR. The Peak Flow Working Group: test of portable peak flow meters by explosive decompression. *Eur Respir J* 1997; 10: Suppl. 24, 23s–25s.
8. Varène P, Vieillefond H, Saumon G, Lafosse JE. Etalonnage des pneumotachographes par méthod intégrale. *Bull Physiopathol Respir* 1974; 10: 349–360.
9. Miller MR, Jones B, Xu Y, Pedersen OF, Quanjer PhH. Peak expiratory flow profiles delivered by pump systems: limitations due to wave action. *Am J Respir Crit Care Med* 2000; 161: 1887–1896.
10. American Thoracic Society. Standardization of spirometry 1994 update (ATS Statement). *Am J Respir Crit Care Med* 1995; 152: 1107–1136.
11. Miller MR, Pedersen OF, Quanjer PhH. The rise and dwell time for peak expiratory flow in patients with and without airflow limitation. *Am J Respir Crit Care Med* 1998; 158: 23–27.