



Comparison of oscillometry devices using active mechanical test loads

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ABSTRACT Noninvasiveness, low cooperation demand and the potential for detailed physiological characterisation have promoted the use of oscillometry in the assessment of lung function. However, concerns have been raised about the comparability of measurement outcomes delivered by the different oscillometry devices. The present study compares the performances of oscillometers in the measurement of mechanical test loads with and without simulated breathing.

Six devices (five were commercially available and one was custom made) were tested with mechanical test loads combining resistors (R), gas compliances (C) and a tube inertance (L), to mimic respiratory resistance (R_{rs}) and reactance (X_{rs}) spectra encountered in clinical practice. A ventilator was used to simulate breathing at tidal volumes of 300 and 700 mL at frequencies of 30 and 15 min^{-1} , respectively. Measurements were evaluated in terms of R , C , L , resonance frequency (f_{res}), reactance area (A_X) and resistance change between 5 and 20 or 19 Hz ($R_{5-20(19)}$).

Increasing test loads caused progressive deviations in R_{rs} and X_{rs} from calculated values at various degrees in the different oscillometers. While mean values of R_{rs} were recovered acceptably, some devices exhibited serious distortions in the frequency dependences of R_{rs} and X_{rs} , leading to large errors in C , L , f_{res} , A_X and $R_{5-20(19)}$. The results were largely independent of the simulated breathing.

Simplistic calibration procedures and mouthpiece corrections, in addition to unknown instrumental and signal processing factors, may be responsible for the large differences in oscillometry measures. Rigorous testing and ongoing harmonisation efforts are necessary to better exploit the diagnostic and scientific potential of oscillometry.



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The clinical utility of oscillometry is limited by the lack of standardisation of devices. This study tested six oscillometers, and reveals very different performances at higher mechanical impedances observed in children and adults with lung disease. <https://bit.ly/317sfjH>

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Introduction

Oscillometry, also known as the forced oscillation technique, has been increasingly employed in the routine assessment of lung function due to the low level of cooperation required from subjects. While the maximum expiratory flow–volume (MEFV) measurement remains the most widespread diagnostic tool in lung function testing, its use is limited to a cooperating age range [1, 2], and it is not feasible as a lung function test in patients with neuromuscular disease [1]. In contrast, high success rates in oscillometry have been reported for both preschool-age children as young as 2 years of age [2] and very elderly subjects who are unable to perform acceptable MEFV manoeuvres [3, 4].

The potential of oscillometry associated with a detailed physiological interpretation of results and a low cooperation requirement have resulted in the development of commercial oscillometry devices replacing various custom-made setups. Standardisation of this technique has become an important task, and recommendations on measurement conditions and accuracy have been published [2, 5, 6]. Since the devices differ in several aspects, such as measurement duration, frequency content, waveform and intensity of oscillations, and signal processing techniques, concerns have been raised about their compatibility in the measurement of respiratory impedance (Z_{rs}). This is particularly important, as numerous databases are being established for various populations with different age groups and ethnicity [7], and the derived reference values would be impractical if they remain device specific. While rigorous criteria and testing procedures have been established for spirometry [8–10], this stage has not been reached in oscillometry.

The importance of device compatibility has long been recognised, and initial studies comparing the performance of two or more oscillometry devices have been conducted in the same subjects [11–19], different study populations [20], or known mechanical test loads [14, 16–18, 21, 22]. However, to our knowledge, reports comparing outputs of all five of the currently marketed oscillometry devices have not been published. The purpose of this study is to fill this knowledge gap by conducting a systematic comparison of outputs from each of the five devices and a custom-made oscillometry setup, by using well characterised mechanical test loads that represent the range of typical Z_{rs} spectra encountered in clinical practice. As breathing has also been suggested as a potential contributor to the diversity in device performances [20] and has been tested in a subset of devices [14], our test loads were measured with and without superimposed simulated breathing signals.

Methods

Study location and equipment

The study was conducted in a single centre (Oscillometry Unit, Centre for Innovative Medicine, McGill University Health Centre Research Institute, Montreal, Canada) using the MasterScreen IOS (Vyaire, Lakeforest, IL, USA), the Quark i2m (Cosmed srl, Rome, Italy), the MostGraph-02 (Chest MI, Tokyo, Japan), the TremoFlo C-100 (Thorasy Medical Systems Inc., Montreal, Canada) and the Resmon Pro (Restech srl, Milan, Italy). A custom-made device based on a wave tube head (length: 17 cm, internal diameter (ID): 1.4 cm) designed for the measurement of high Z_{rs} values [23, 24] was also tested.

Mechanical test loads

Six different mechanical test loads were assembled (figure 1) using five different stacked mesh resistors (1.8, 4.5, 8.4, 11.8 and 14.0 hPa·s·L⁻¹ at the unidirectional (dc) flow of 50 mL·s⁻¹) simulating respiratory resistance (R_1 , R_2 , R_3 , R_4 and R_5 , respectively), a single 30.5-cm PVC tube (1.9 cm ID) representing inertance (L), and two different elastic loads composed of 4- and 23-L glass bottles as compliances (C). A side port in each of the bottles was connected to a large animal ventilator (model 613; Harvard Apparatus, Holliston, MA) delivering sinusoidal volume changes *via* a 100- and 200-cm 3.2-mm-ID polyethylene tube in the case of the 23-L bottle (C_1) and the 4-L bottle (C_2), respectively. The purpose of this tubing was to add high impedance between the test loads and the ventilator to prevent the dynamic contribution of the

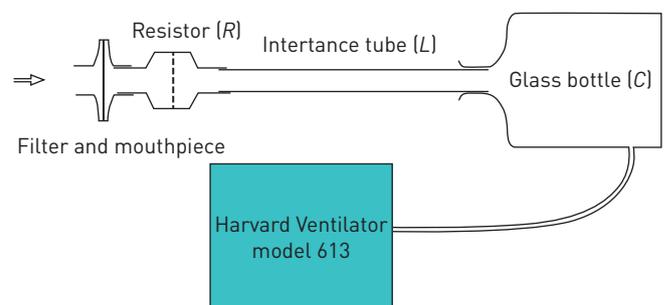


FIGURE 1 Schematic arrangement of the mechanical test loads.

TABLE 1 Mechanical test loads (M1–M6) as combinations of resistors and compliances (the same inertance tube was used in all test loads) and ventilation parameters

| Mechanical test load | M1 | M2 | M3 | M4 | M5 | M6 |
|--|--|--|--|-----|--|--|
| Resistor (R) $\text{hPa}\cdot\text{s}\cdot\text{L}^{-1}$ | R_1 1.8 [#] /1.9 [¶] | R_2 4.5 [#] /4.8 [¶] | R_3 8.4 [#] /8.7 [¶] | | R_4 11.8 [#] /12.4 [¶] | R_5 14.0 [#] /14.2 [¶] |
| Compliance (C) $\text{mL}\cdot\text{hPa}^{-1}$ | | C_1 16.80 | | | C_2 2.95 | |
| Tidal volume mL | 700 | 700 | 700 | 300 | 300 | 300 |
| Frequency min^{-1} | 15 | 15 | 15 | 30 | 30 | 30 |

[#]: measured at $50 \text{ mL}\cdot\text{s}^{-1}$; [¶]: measured at the mean flow of the ventilator setting.

variable pump volume to C . The test model combinations and the ventilation parameters are shown in table 1. The signals recorded during the oscillatory measurements are shown in figure 2.

The input impedance of the test loads (Z_m) was calculated in the frequency range 4–38 Hz as follows. The input impedance of the inertance tube and the gas bottle (Z_{LC}), attached as the load impedance of the tube, were computed from the numerical solutions of the Navier–Stokes equations as cylindrical conduits with an open and closed end for the tube and the bottle, respectively [25]. The impedance of the resistor was modelled by a lumped T-network consisting of the in-series tube impedances and the parallel shunt impedances of the gas volume in front of and beyond the metal mesh screen assembly, all computed as above [25], while the resistance value of the mesh screen was taken from the dc flow measurements made at the mean flow of the actual ventilator setting (table 1). This T-network was loaded by the Z_{LC} to obtain Z_m . The impedance of the bacterial filter and mouthpiece assembly was accounted for by the specific calibration procedure employed in each device.

The calculated Z_m values of these test loads deviated mildly from an ideal R – L – C impedance, due to the following mechanisms: 1) The gas in the glass bottle undergoes a polytropic process implying a slight difference from the strictly hyperbolic course of the imaginary part and the presence of a minor real part at the lowest frequencies; the latter was observable in the case of the smaller bottle as a small drop in test load resistance (R_m) of $0.06 \text{ hPa}\cdot\text{s}\cdot\text{L}^{-1}$ between 4 and 10 Hz. 2) An opposite change in R_m ($0.13 \text{ hPa}\cdot\text{s}\cdot\text{L}^{-1}$ between 10 and 38 Hz) was caused by the positive frequency dependence of R in the inertance tube, as a consequence of the distortion of the parabolic velocity profile [26]. All these departures resulted in small (0.3–1.2%) errors in the R – L – C fitting to the computed impedance spectra.

Oscillation frequencies

The tested devices employed different oscillation signals (table 2), with frequency values ranging from 5 Hz to 35–37 Hz, except in the Quark i2m device whose lowest frequency was 4 Hz and highest was 48 Hz. In order to calculate the reactance area (A_X) between the lowest frequency and the resonance frequency (f_{res}) uniformly, the values of reactance (X_m) at 5 Hz were interpolated from the 4-Hz and 6-Hz data. If f_{res} was beyond the measurement frequencies, integration was truncated at the highest frequency.

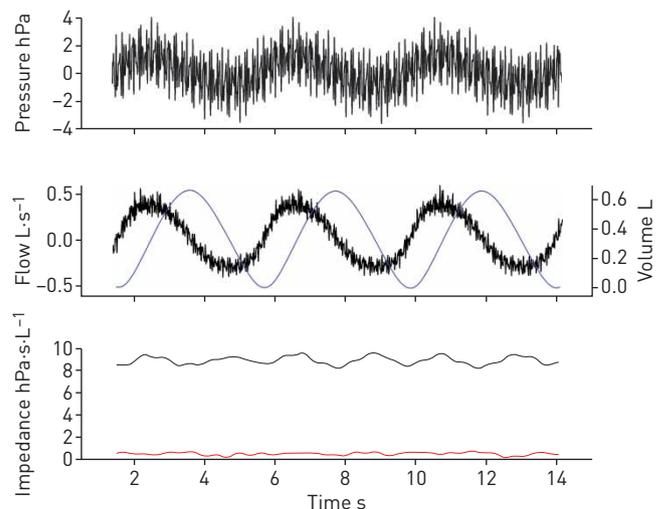


FIGURE 2 Representative tracing of pressure, flow and volume (blue) signals with “intra-breath” resistance (black) and reactance (red) computed at an 11-Hz oscillation frequency in test load M2.

TABLE 2 Types and frequency contents of the oscillation signals

| Device | Frequencies Hz | Signal type | Recording time |
|------------------|--------------------------------------|--------------------------------|----------------|
| Wave Tube | 5, 11, 13, 17, 19, 23, 29, 31, 37 | Pseudo-random, relative primes | 16 s |
| TremoFlo C-100 | 5, 11, 13, 17, 19, 23, 29, 31, 37 | Pseudo-random, relative primes | 16 s |
| MasterScreen IOS | 5, 10, 15, 20, 25, 35 | Recurrent impulses | 30 s |
| MostGraph-02 imp | 5, 10, 15, 20, 25, 30, 35 | Recurrent impulses | 60 s |
| MostGraph-02 prn | 5, 10, 15, 20, 25, 30, 35 | Pseudo-random | 60 s |
| Quark i2m | 4, 6, 8, ..., 32, 34, 36 | Pseudo-random | 8 s |
| Resmon Pro | 5, 7, 11, 13, 17, 19, 23, 29, 31, 37 | Pseudo-random, relative primes | 30 s |
| Resmon Pro 3f | 5, 11, 19 | Pseudo-random, relative primes | 5 breaths |

A commonly used measure of peripheral respiratory mechanics, the difference in R_{rs} between 5 and 20 or 19 Hz (whichever is available) was also characterised by the $R_{5-20(19)}$ values. Two devices (MostGraph-02 and Resmon Pro) each offered two different spectral signal types (table 2), and the tests were accomplished with both versions. The amplitudes of the oscillation signals were set as those implemented or recommended by the manufacturers. With each test model and device, five measurements were made and the impedance spectra were ensemble averaged.

Statistical analysis

The effects of simulated breathing on the estimates of R , L , C , f_{res} , A_X and $R_{5-20(19)}$ were analysed with two-way repeated measures (RM) ANOVA (see details in supplementary appendix S1 and tables S1–S2).

Results

Measurements with the Resmon Pro device are reported for test models M1–M3 and M1–M4 with the 10-component and the three-component test signals, respectively, in accordance with the accuracy limits in impedance specified by the manufacturer. Impedance spectra obtained on the test models are illustrated with the lowest-impedance (M1) and highest-impedance model (M6) measurements with the different devices and signal modes in figure 3. At the lowest test load of M1, most devices measured slightly higher values of resistance and reactance compared to the calculated values; however, apart from a single slightly differing device (MostGraph) the performances were similar (figure 3a), whereas with the highest test load of M6, dramatic differences developed between devices (figure 3b). The results for all models of M1 through to M6, with or without simulated breathing are shown in figure S1a–f. With increasing values of

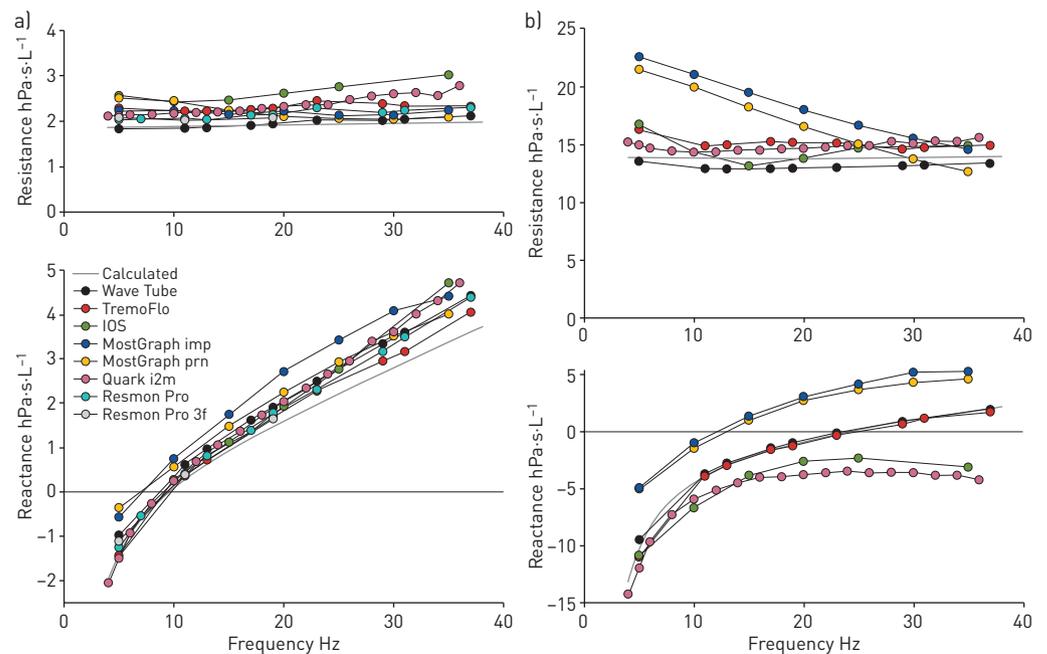


FIGURE 3 Impedance measurements in the mechanical test loads a) M1 and b) M6.

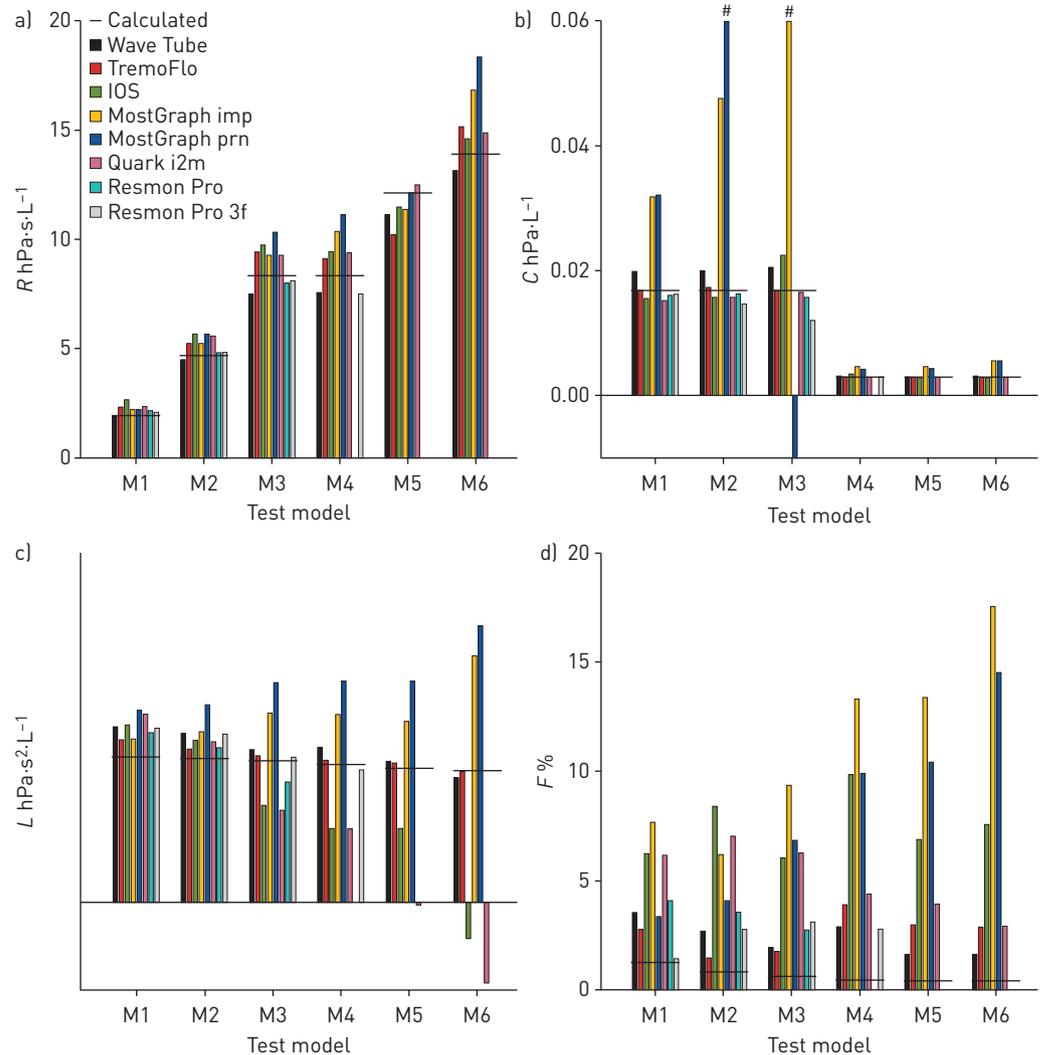


FIGURE 4 Values of a) resistance (R), b) compliance (C), c) inertance (L) and d) the fitting error (F) obtained from model fitting to impedance data measured in the mechanical test loads M1–M6, with the different devices and modes. Horizontal lines indicate values from fitting to calculated impedances. #: $C > 0.06 \text{ L} \cdot \text{hPa}^{-1}$.

resistors but invariable C and L elements in the test loads (M1, M2 and M3), the different frequency dependences in resistance become enhanced and the courses of reactance deviated significantly from the calculated values in half of the devices (IOS, i2m and MostGraph). At the higher resistive and elastic values of the test loads M4, M5 and M6, most devices yielded similar resistance spectra but this was associated with reactance courses diverging further from the predictions in the same three devices. Apart from the slight elevation in resistance, also reflected by the calculated values, there was no systematic difference in impedance between the static and dynamic test model measurements (*i.e.* with and without simulated breathing with any of the devices) (figure S1a–f).

Figures 4 and 5 illustrate the values of Z_m measures for all test loads and devices. Resistance, R obtained as the average value of R_m in the measured frequency range agreed fairly well with the calculated dc resistances by all devices (figure 4a), with the largest deviation exhibited by the MostGraph device at the highest test load. The theoretical values of C were accurately recovered at all test loads by all devices except the MostGraph, where the systematic underestimation of the elastic impedance led to unrealistic high values of C (figure 4b), in accordance with the spectral results shown in figures 3 and S1a–f. The markedly different high-frequency behaviour of the different devices is sensitively reflected by the values of L , which greatly overestimated (MostGraph) or underestimated (IOS and i2M) the theoretical values progressively with increasing test loads (figure 4c). While the impedance spectra calculated for the test loads were fairly consistent with the ideal R – C – L behaviour, as manifested in fitting errors (F) mostly $< 1\%$, the spectral distortions introduced by some devices led to enormously high F values (figure 4d).

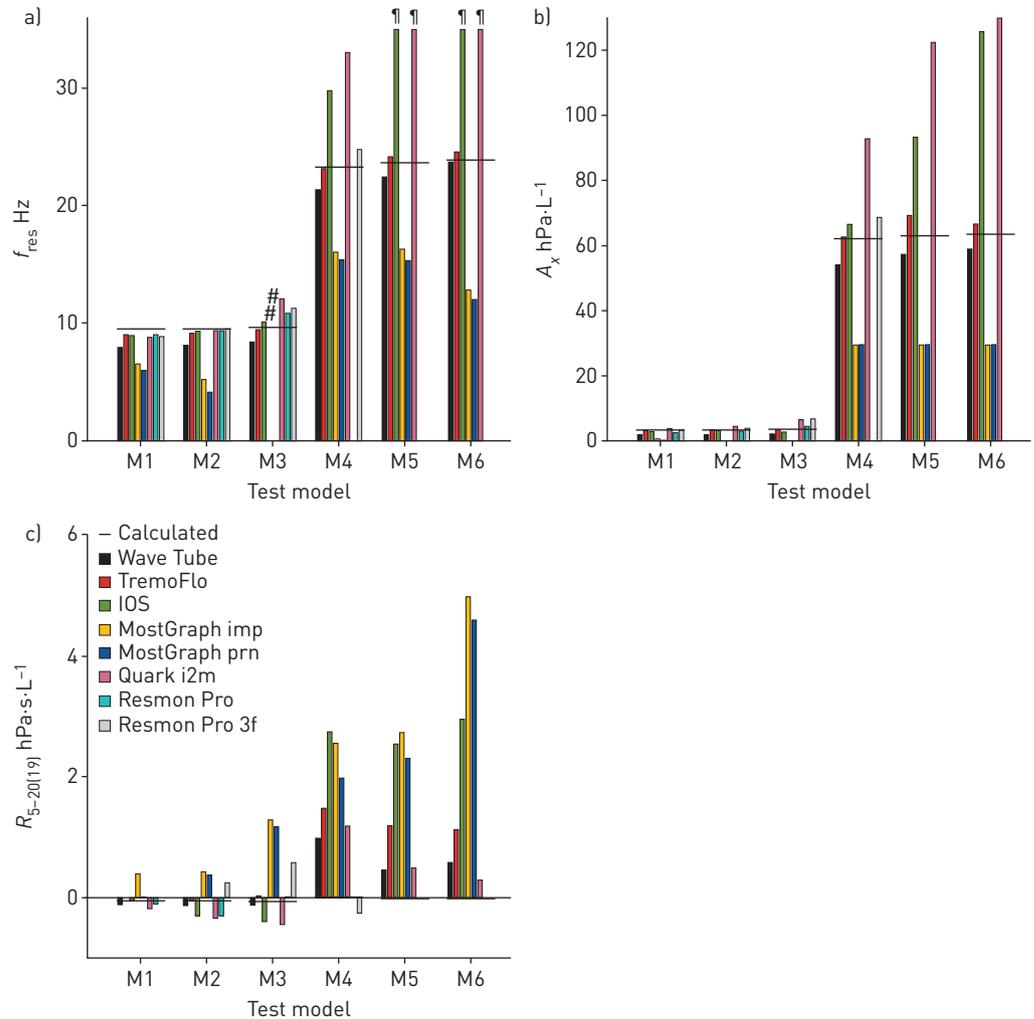


FIGURE 5 Values of a) resonance frequency (f_{res}), b) reactance area (A_x) and c) the frequency dependence of resistance ($R_{5-20(19)}$) calculated from impedance data measured in the mechanical test loads M1–M6 with the different devices and modes. Horizontal lines indicate values obtained from the calculated impedance data. Zero crossing of reactance occurred below ($\#$) or above the measured frequency range (\dagger).

Device performances were also characterised in terms of commonly employed Z_{rs} measures, such as f_{res} , A_x and $R_{5-20(19)}$ (figure 5). The theoretical values of f_{res} were nicely recovered by some devices (Wave Tube, TremoFlo and Resmon Pro), while they were markedly mis-estimated or even undetected by the rest of the devices (figure 5a). Inaccurate estimation of f_{res} led to large errors in the values of A_x , most remarkably in the low-compliance test models (figure 5b). Although there was negligible frequency dependence of resistance in all test models, artifactually high values of $R_{5-20(19)}$ were observed with some devices (figure 5c).

In accordance with the display of Z_m spectra (figure S1a–f) the two-way RM ANOVA revealed systematic but minor changes in R caused by the simulated breathing; the rest of the impedance parameters remained unaffected (tables S1 and S2).

Discussion

The present work is aimed at promoting the ‘standardisation of oscillometry’, an action that was initiated for spirometry in 1979 [27] and was completed in 2005 [9]. This study uniquely benefits from the open and consensual attitude toward the comparative study, expressed by the manufacturers of the commercial oscillometry devices, to solve the standardisation problems exposed 25 years ago [6, 22].

The presented results reveal substantial differences in the estimation of impedances of known mechanical models of the respiratory system and highlight the limitations in the compatibility of oscillometry devices. A typical finding was that the deviations of measured impedance increased with the magnitude of the test load in

some devices, and these manifested dominantly in the course of the reactance values. While almost all devices were relatively accurate in measuring reactance at the lower frequencies, poor performance at higher frequencies resulted in widely variable values in f_{res} and A_X , which are common indices to describe the elasticity of the respiratory system and the inhomogeneity of the lung periphery [5, 28, 29]. Smaller differences were seen in the courses of R_m ; however, these were magnified in the scattering values of $R_{5-20(19)}$, a popular measure suggested to characterise peripheral airway obstruction [29]. Although the physiological validity of these oscillometry measures has yet to be confirmed, the huge biases due to the different performances of the oscillometry devices reduce their utility in clinical practice; indeed, interdevice comparisons [16–19] have raised concerns about systematic differences in these measures. These findings also emphasise that standardisation criteria [5] based on single values of Z_{rs} (or R_{rs} , X_{rs} and $|Z_{rs}|$) alone are insufficient because of the growing importance of the derived indices mentioned above.

Due to the lack of public information on the hardware and software details of each device, including the linearity and common-mode behaviour of the flow meters, and the details of signal processing, the reasons underlying the diverse performances cannot be elucidated. Concerns have been raised about the low-impedance calibration devices used in some equipment despite methodological recommendations published long ago [5, 6], which are far too small compared to the Z_{rs} values encountered in clinical practice. Another concern relates to the simplistic correction for the filter–mouthpiece assembly, which does not take into account the shunt effect of the associated dead spaces on the measured Z_{rs} [15, 30].

A limitation of the current study is the inclusion of empirical values of R in the test load assembly, in the absence of theoretical calculations for the stacked mesh resistors. Measurements of R with dc flow revealed slight nonlinearity (data not shown). Indeed, as exemplified in figure 2 by the flow-dependent fluctuations in R (and X), the resistive component of the test loads did not behave ideally (this might also have contributed to the differences between the oscillometry devices delivering different factory-set signal amplitudes). Apart from this uncertainty in the actual values of R (and to a lesser extent, of L) the rheology of the test load impedances was fairly consistent with the R-C-L model whose parameters sensitively reflected the differences in the performance of the oscillometry devices.

Finally, some aspects of the device–subject interaction should be mentioned here. In an earlier comparative study in five groups of healthy subjects measured with five different oscillometry devices [20], the intergroup differences in Z_{rs} were larger than expected on the basis of the fairly uniform recovery of the test load impedance of $10 \text{ hPa}\cdot\text{s}\cdot\text{L}^{-1}$, suggesting an altered performance of the devices during respiratory flow. The present study does not confirm this suggestion, as simulated breathing introduced unimportant differences in the frequency dependence of Z_{rs} and the model parameters, although the use of a sinusoidal simulated breathing was a methodological restriction. Equipment impedance, dead space, oscillation waveform and measurement duration may change Z_{rs} *via* alteration of the breathing pattern [14]. Although this was not addressed in the present study, it is unlikely that these factors would have an effect comparable to the distortions observed in the frequency responses of the different devices. It is worthwhile to note that one device (MostGraph-02) was tested in both impulse and pseudo-random signal modes and the results were basically similar. The current study does not address the marked nonlinearities present in the upper airways and flow-limiting bronchial segments, causing large intra-breath fluctuations in R_{rs} and X_{rs} [31–35], a factor potentially affecting the device–subject interaction and hence the device performance. However, new modalities of oscillometry that widen the utility of this method in lung function testing [33, 34] and monitoring [36], impose further requirements over the accuracy of mean R_{rs} and X_{rs} .

In conclusion, the present testing of commercial oscillometry devices reveals progressive differences in measurement performance with increasing impedance values, which exclude the comparability of results from studies using different equipment and prohibit the formation of large device-independent oscillometry databases. Harmonisation, including the use of rigorous testing procedures, and, if indicated, revision of technical solutions in individual equipment types are necessary in order to exploit the diagnostic potential of oscillometry in the lung function arsenal. The consensus in communicating the results of this comparative study expresses the recognition of this urgent need by all manufacturers and researchers involved.

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Conflict of interest: R.J. Dandurand reports that he is a 15% shareholder in SpiroTech Medical Inc., which holds the patent for a novel device for the home monitoring of respiratory system resistance; and unrestricted educational grants from AstraZeneca, Boehringer Ingelheim, Novartis, Pfizer and Teva Pharma, outside the submitted work. J-P. Lavoie has nothing to disclose. L.C. Lands has nothing to disclose. Z. Hantos reports that he is named as an inventor on a patent owned by the Telethon Kids Institute entitled “A method of diagnosing a respiratory disease or disorder or monitoring treatment of same and a device for use therein” (Australian patent application number 2005903034). The techniques used in this study are broadly consistent with this patent. He receives no royalties from, nor has he any royalty agreement with, the Telethon Kids Institute under this patent. He also has a consultancy agreement with Thorasys Medical Systems, Inc., which is unrelated to the subject of the present study and was established after the present study was performed.

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