Automating the correction of flow integration drift during whole-body plethysmography

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***Abstract*— Prolonged measurement of total body volume variations (deltaVb) with whole-body, flow-based plethysmography (WBP) results in a drift of the signal due to changes in temperature and humidity inside the plethysmograph and to numerical integration of the flow to obtain deltaVb. This drift has been previously corrected with the application of a wavelet-based filter using visual inspection of the signal to select the optimal filter level (Uva et al. *Front. Physiol.* 6:411, 2016), thus introducing potential operator bias. To exclude the latter we compared this approach with a newly developed automatic method based on (1) correction for actual changes in temperature and humidity inside the plethysmograph (algorithm TH) and (2) automatic selection of the wavelet filter level based on comparison between deltaVb and intra-thoracic and abdominal pressure variations measured simultaneously (algorithm WAV). The Pearson's correlation coefficient between deltaVb and the changes in volume of the chest wall (deltaVcw) simultaneously obtained by optoelectronic plethysmography (OEP) was calculated after correction of deltaVb with TH and WAV applied separately, TH and WAV applied consecutively (TH+WAV), manual selection of a wavelet filter based on visual inspection (MAN) or no correction (CTRL). The correlation between deltaVb and deltaVcw increased marginally with WAV, TH+WAV and MAN compared to CTRL ($P < 0.01$). Conversely, TH alone yielded a lower correlation ($P < 0.01$). It follows that while the automated wavelet filter level selection method (WAV) represents an effective, operator-independent method for the correction of deltaVb, whether or not it is combined with specific correction for changes in thermodynamic conditions inside the plethysmograph, the manual method (MAN) yields satisfactory results without the constraints of intra-thoracic and abdominal pressure measurement.**

I. INTRODUCTION

Double body plethysmography is a novel technique allowing to quantify blood displacements between the trunk and the extremities. It combines the measurements of total body volume variations (ΔVb) using classical flow whole-body plethysmography (WBP) with chest wall volume variations (ΔVcw) using optoelectronic plethysmography (OEP) [1].

ΔVb represents changes in lung volume due to airflow and gas compression or decompression, while ΔVcw is affected by the same variables, plus or minus the volume of blood displaced between the trunk and the extremities (Vbs). Therefore, Vbs is obtained by subtracting the two signals acquired simultaneously.

Since ΔVb is measured using a flow plethysmograph, the signal needs to be corrected beforehand for a nonlinear drift resulting from (1) changes in thermodynamic conditions inside the plethysmograph due to heat and water vapor production by the participant and (2) numerical integration of the flow signal to obtain a volume. While the drift can be considered negligible over a short period of time, it must be corrected during prolonged monitoring of ΔVb [2]–[5]. This is especially true during exercise, when temperature and humidity inside the cabin increase more rapidly. Uva et al [6] used wavelet filtering to correct for this nonlinear drift, relying on visual inspection of the signal to select the optimal filter. While yielding satisfactory results, this method introduces a potential operator bias.

In order to exclude operator bias we therefore evaluated automating of signal processing including correction based on actual changes in temperature and humidity inside the plethysmograph and correspondence with independently measured physiological variables.

II. METHODS

A. Data collection

The files analyzed were collected during an experiment conducted with ten healthy participants (25 ± 3 y, 177 ± 5 cm, 67.4 ± 5.8 kg, mean ± SD) performing transitions from rest to cycling exercise inside the plethysmograph (ethics approval was obtained from the local IRB). Prior to data collection, two balloon-tipped catheters connected to pressure transducers (RCEM250DB, Sensortechnics, Germany) were introduced nasally into the participant's esophagus and stomach for continuous measurement of esophageal (Pes) and abdominal (Pab) pressure, respectively. 89 retro-reflective markers were taped on the participant's thorax for the measurement of ΔVcw using an optoelectronic system (OEP, BTS Bioengineering, Italy). Each participant performed nine transitions after 1 minute of rest to 3 minutes of cycling exercise at 100 W, before recovering for 2 minutes. They breathed through a mouthpiece connected to room air through the plethysmograph's wall and wore a nose-clip. Eight Peltier cells combined with a set of fans placed at the bottom of the cabin minimized changes in temperature and humidity avoiding condensation on the walls of the plethysmograph. Between trials (i.e. every 5 min), the WBP was opened in order to restore initial thermodynamic conditions.

B. ΔVb signal correction

The correction method consists of two algorithms applied consecutively: (1) an algorithm to correct the drift due to changes in thermodynamic conditions and (2) an algorithm to automatically select the level of a wavelet filter applied

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to remove the remaining drift mainly due to numerical integration of the flow signal to obtain ΔVb.

C. Correction for changes in thermodynamic conditions

During prolonged WBP measurements, especially when performing exercise, the rise in temperature and humidity inside the plethysmograph lead to a change in pressure, generating air flow out of the plethysmograph. This additional flow is superimposed on the cyclical flow generated by the participant's variations in body volume due to breathing. Thus, integration of the flow signal comprises an additional component and does not truly reflect ΔVb. Our algorithm uses the laws of thermodynamics to continuously compute this extra volume and subtract it from ΔVb.

The variation in volume due to changes in thermodynamic conditions (ΔVtc) between time i and i+1 can be written as:

$$\Delta Vtc(i) = G(i) \cdot Vair \tag{1}$$

where Vair is a constant calculated as the difference between the total volume of the plethysmograph and the volume occupied by the participant's body – the latter being estimated by expressing body mass in liter – and G represents the increase in volume due to changes in thermodynamic conditions. G can be written as a function of the relative humidity and temperature inside the plethysmograph at time i and i+1:

$$G(i) = \left(\frac{Patm - (H(i) \cdot Pv(i))}{Patm - (H(i+1) \cdot Pv(i+1))} \cdot \frac{T(i+1)}{T(i)}\right) - 1 \tag{2}$$

where Patm is the atmospheric pressure, Pv the vapor pressure, T the temperature expressed in kelvins and H the relative humidity expressed between 0 and 1. H and T were measured continuously with a digital sensor (DHT22, Aosong, China) placed inside the plethysmograph. Pv was calculated for each sample of T (expressed in °C) as:

$$Pv(i) = 610.5 \cdot e^{\frac{17.269 \cdot T(i)}{237.3 + T(i)}} \tag{3}$$

Vtc was computed sample by sample and then subtracted to ΔVb to correct the thermal drift. The integral ΔVtc in sample domain (Vtc) is the volume increase signal that is superimposed on ΔVb because of the changes in thermodynamic conditions and can be subtracted from ΔVb to correct the thermal drift:

$$\int_i \Delta Vtc(i)\, dt = \int_i G(i) \cdot Vair\, dt \tag{4}$$

D. Wavelet filtering

As the TH algorithm was designed to correct the ΔVb signal solely for changes in temperature and humidity, the remaining noise was corrected by subsequently applying a wavelet filter. As previously described by Uva et al [6], we used a second order discrete biorthogonal function as the mother wavelet. We developed a second algorithm to automatically select the optimal level of the wavelet filter, using the correspondence between independently measured physiological parameters as a reference. Since Vbs variations strongly depend on intra-abdominal (Pab) and intra-thoracic (esophageal, Pes) pressure variations [1] the principle of this algorithm was to maximize the correlation between the overall variations of blood shift ($V_{bs,n}$) obtained by the difference ΔVcw and the variation of wavelet-filtered Vb at level n ($\Delta V_{b,n}$), Pab and Pes:

$$max_n[\rho(V_{bs,n}, P_{ab}) + \rho(V_{bs,n}, P_{es})] \tag{5}$$

where the volume of blood shift can be expressed as the difference between the chest wall volume variations and the filtered box volume at level n:

$$V_{bs,n} = \Delta V_{cw} - \Delta V_{b,n} \tag{6}$$

Substituting Eq. (6) into Eq. (5):

$$\begin{aligned} &max_n[\rho(V_{bs,n}, P_{ab}) + \rho(V_{bs,n}, P_{es})] \\ &\quad = max_n[\rho(\Delta V_{cw} - \Delta V_{b,n}, P_{ab}) \\ &\quad + \rho(\Delta V_{cw} - \Delta V_{b,n}, P_{es})] \end{aligned} \tag{7}$$

Considering that the wavelet decomposition was used to create a high-pass filter, the filtered ΔVb signal can be expressed as raw Vb minus its wavelet decomposition at level n (Vn):

$$\Delta V_{b,n} = \Delta V_{b,0} - V_n \tag{8}$$

Substituting Eq. (8) into Eq. (7) and applying the distributive property of the correlation function, Eq. (5) can be rearranged as:

$$\begin{aligned} &max_n[\rho(\Delta V_{cw} - \Delta V_{b,0}, P_{ab}) + \rho(\Delta V_{cw} - \Delta V_{b,0}, P_{es}) \\ &\quad + \rho(V_n, P_{ab}) + \rho(V_n, P_{es})] \end{aligned} \tag{9}$$

and thus Eq. (9) can be rewritten as:

$$\begin{aligned} &max_n[\rho(V_{bs,n}, P_{ab}) + \rho(V_{bs,n}, P_{es})] \\ &= max_n[\rho(V_n, P_{ab}) + \rho(V_n, P_{es})] + k \end{aligned} \tag{10}$$

where k is a constant.

Therefore, the principle of this algorithm is to find the wavelet level that maximizes the correlation between the pressure signals and the residual extracted from the wavelet filtered ΔVb which, as demonstrated, is equivalent to maximizing the correlation between the pressure signals and Vbs. As Aliverti et al [1] reported a correspondence between the pressure signals and Vbs in their overall shapes, we extracted the overall signal variations of Pab, Pes, by applying a Butterworth low-pass filter with a cut-off frequency fixed at half the breathing frequency at rest, prior to compute the correlations. Pearson's correlation coefficients between pressures and Vn were calculated on windows of variable sizes, determined by the inflection points of each filtered pressure signal (Fig. 1). Within each window, the maximal value between $\rho(Vn, Pab)$ and $\rho(Vn, Pes)$ was extracted (ρmax). The algorithm output was the wavelet level that maximized the sum of every ρmax over the whole signal.

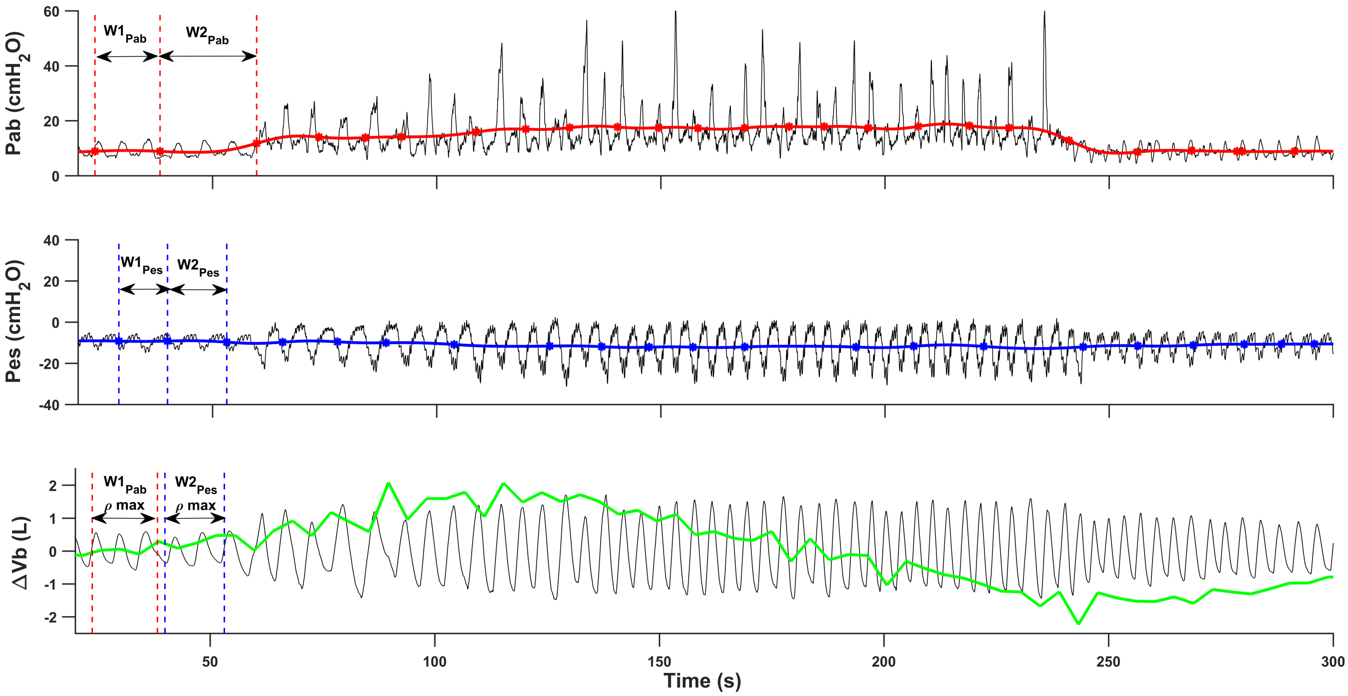

Fig. 1. From top to bottom, tracings of abdominal pressure (Pab), esophageal pressure (Pes) and total body volume variations (ΔVb). Signals are displayed in black with overlapped Pab filtered (red), Pes filtered (blue) and raw ΔVb minus its wavelet decomposition at level n (V_n) (green), respectively. Dashed intervals identify the selected windows on Pab and Pes, respectively. For each window, the maximum value (ρmax) between $\rho(Vn, Pab)$ and $\rho(Vn, Pes)$ was taken into account.

E. Data analysis

In order to assess the validity of our method, we compared the corrected ΔVb signal with the ΔVcw signal obtained with OEP, as it is an optical method that accurately reflects changes in lung volume [7]. A Pearson's correlation between Vcw and Vb was computed for every file with the signals corrected using TH and WAV separately, both algorithms applied consecutively (TH+WAV), manual selection of the wavelet filter level (MAN) or no correction (CTRL). A one-way repeated measure ANOVA with Tukey's correction for multiple comparisons was performed to detect significant changes between conditions. Significance was set at $P < 0.05$.

III. Results

Our results indicate that WAV, TH+WAV and MAN produced an incremental increase in $\rho(\Delta Vb, \Delta Vcw)$ (0.88 ± 0.05 vs 0.90 ± 0.04 vs 0.93 ± 0.05, respectively) compared to CTRL (0.74 ± 0.07) ($P < 0.01$). Conversely, TH alone yielded a lower $\rho(\Delta Vb, \Delta Vcw)$ (0.39 ± 0.11) compared to all conditions (Fig. 2 and 3).

IV. Discussion

The aim of this study was to automatize the correction of ΔVb during prolonged measurement periods and compare the results obtained with the previously described manual method. Our results indicate that the combination of TH and WAV increased $\rho(\Delta Vb, \Delta Vcw)$ to the same extent as the previously described manual method (MAN) [6] compared to CTRL (Fig. 2 and 3). Applying the second algorithm (WAV) directly on the raw ΔVb signal produced a similar improvement. Thus, correcting ΔVb for changes in

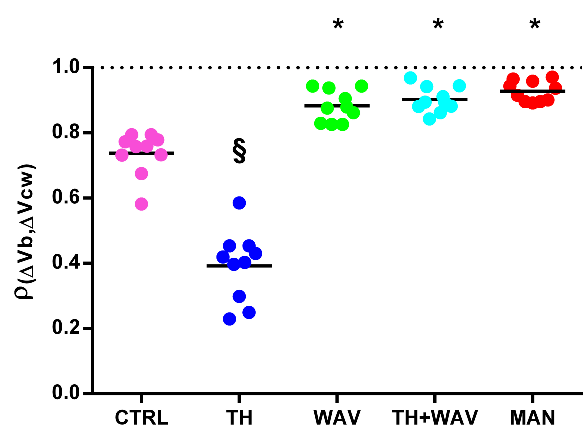

Fig. 2. Pearson's correlation coefficient between ΔVb and ΔVcw ($\rho(\Delta Vb, \Delta Vcw)$), with no correction of ΔVb (CTRL), correction based on thermodynamic changes (TH), correction based on wavelet filtering with automatic level selection (WAV), both TH and WAV applied sequentially (TH+WAV) or manual selection of the wavelet filter level based on visual inspection of the signal (MAN). *Significant increase compared to CTRL; §Significant decrease compared to CTRL; $P < 0.01$.

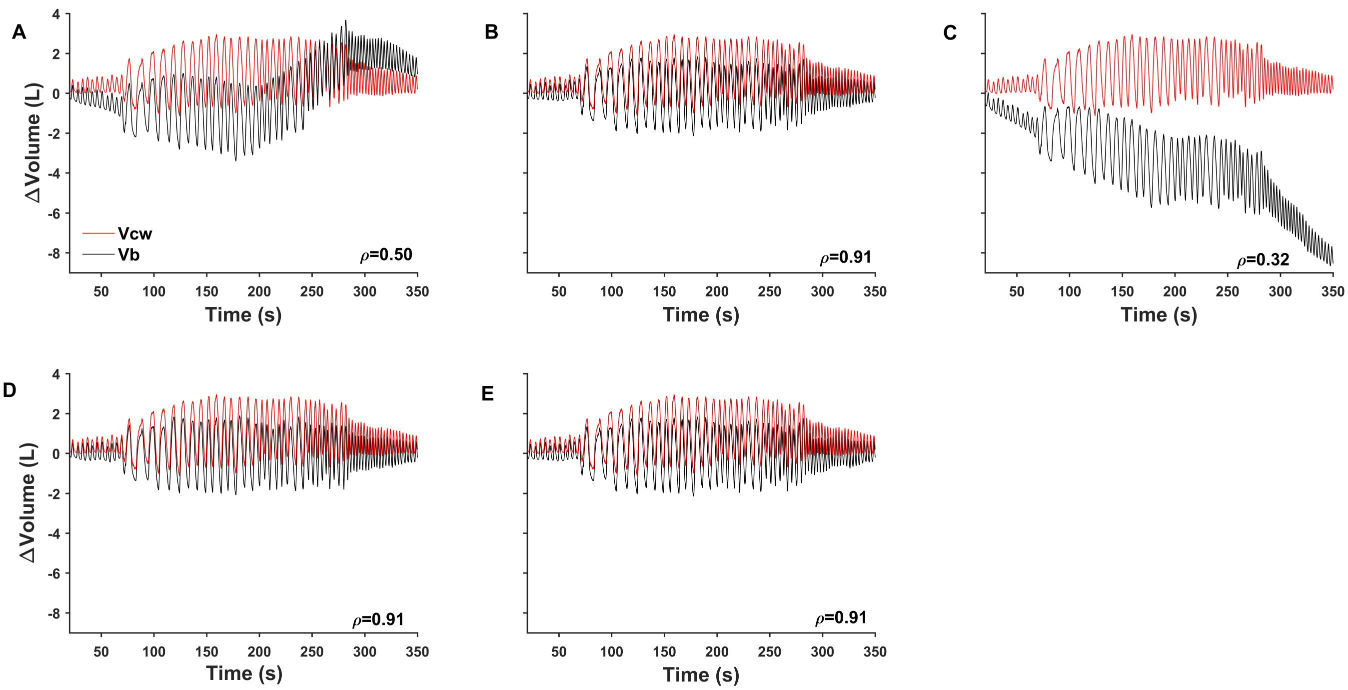

Fig. 3. Different methods of ΔVb correction. *A*: ΔVcw in red with overlapped ΔVb with no correction (CTRL), *B*: correction based on wavelet filtering with automatic level selection (WAV), *C:* correction based on thermodynamic changes (TH), *D:* both TH and WAV applied consecutively (TH+WAV), *E:* manual selection of the wavelet filter level based on visual inspection of the signal (MAN).

thermodynamic conditions inside the WBP cabin appeared superfluous, suggesting that the drift due to mathematical integration of the flow signal is the main cause for the distortion of the ΔVb signal. It follows that WAV *per se* is sufficient to correct ΔVb satisfactorily, thus indicating that temperature and humidity monitoring are not required.

These results show that the automatic selection of the wavelet filter level was successful in correcting ΔVb without the need of manual operations, which alleviates any potential operator bias. However, it should be noted that this method requires simultaneous measurement of Pab and Pes. Monitoring thoraco-abdominal pressure changes is mandatory when studying interactions between pulmonary and circulatory adjustments, justifying their invasive nature. As our results showed that MAN yielded similar results compared to WAV, it can be argued that MAN represents a satisfactory method to correct ΔVb when the research question does not impose monitoring of Pab and Pes.

V. Conclusions

Manual selection based on visual inspection of the signal (MAN) yields similar results as the proposed automatic wavelet filter level selection method (WAV), whether or not it was combined with specific correction for changes in thermodynamic conditions (TH). Therefore, while the proposed pressure-guided automated WAV represents a valuable strategy to exclude potential operator bias when Pes and Pab are monitored anyway for specific research questions, the manual selection method can be used in a more broad range of settings that do not require pressure monitoring. These results offer promising perspectives towards the use of double body plethysmography during prolonged measurement periods as well as during exercise.

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