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Detecting variations of blood volume shift due to heart beat from respiratory inductive plethysmography measurements in man

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Abstract

The simultaneous study of the cardiac and respiratory activities and their interactions is of great physiological and clinical interest. For this purpose, we want to investigate if respiratory inductive plethysmography (RIP) can be used for cardiac functional exploration. We propose a system, based on RIP technology and time-scale approaches of signal processing, for the extraction of cardiac information. This study focuses on the monitoring of blood volume shift due to heart beat, noted $\Delta V_{r-c}$ and investigates RIP for the detection of $\Delta V_{r-c}$ variations by comparison to stroke volume (SV) variations estimated by impedance cardiography (IMP). We proposed a specific respiratory protocol assumed to induce significant variations of the SV. Fifteen healthy volunteers in the seated and supine positions were asked to alternate rest respiration and maneuvers, consisting in blowing into a manometer. A multi-step treatment including a variant of empirical mode decomposition was applied on RIP signals to extract cardiac volume signals and estimate beat-to-beat $\Delta V_{r-c}$. These were averaged in quasi-stationary states at rest and during the respiratory maneuvers, and analysed in view of SV estimations from IMP signals simultaneously acquired. Correlation and statistical tests over the data show that RIP can be used to detect variations of the cardiac blood shift in healthy young subjects.

Keywords: respiratory inductive plethysmography, impedance cardiography, stroke volume, empirical mode decomposition, cardio-respiratory exploration

(Some figures may appear in colour only in the online journal)

1. Introduction

Respiratory inductive plethysmography (RIP) has been validated for respiratory rate monitoring (Lymberis and Dittmar 2007, Grossman et al 2010) and used for ventilatory
Figure 1. (a) Computer-assisted RIP vest (Visurespi®, RBI, France). (b) The calibrated signal of thoracic volume variations obtained with the RIP vest. The arrows indicate examples of supposed cardiac events, according to the simultaneous ECG signal.

function assessment (Chadha et al 1982, Tobin et al 1983, Eberhard et al 2001). This non-invasive technology is considered as wearable technology for breathing monitoring (Lanatà et al 2010) in the current context of continuous monitoring of vital behavioural signs, an emerging concept of healthcare. However, while the developed wearable systems are usually dedicated to the monitoring of one specific type of physiological sign, mainly to simplify technologic concerns, the peripheral nature of RIP measurements gives access to a large amount of information due to multiple physiological interactions. Indeed, RIP allows us to measure the variations of the chest volume which includes the lungs and the heart. Therefore, one may suppose that one RIP signal is a linear combination of one respiratory component and one cardiac component. Besides, Aliverti et al (2010), as well as previous authors (Blair and Wedd 1939), have described a cardiac component on the chest volume variations recorded via rather cumbersome plethysmographic tools.

The challenge is to extract and give a sense to the component of small amplitude, supposed cardiac, and discernable (figure 1) in RIP signal, which is mainly respiratory in terms of amplitude. We want to investigate if RIP can be a cardiac functional exploration tool. This would allow us to jointly monitor the cardiac and respiratory activities and study cardiorespiratory (CR) interactions, which are of great physiological and clinical interest (Bradley et al 2010, Lalande and Johnson 2010, Marcora et al 2008).

In this study, we investigate the possibility of extracting cardiac information from RIP signals and then applying RIP technology for cardiac monitoring. Cardiac extraction from RIP signals is the result of a multi-step processing: (1) decomposition and representation of the embedded components, (2) reconstruction of a cardiac volume signal and (3) estimation of cardiac parameters. These steps will be described in detail in section 3.

The first step is the differentiated representation of the cardiac and respiratory components entangled in the RIP signal, with respect to their interactions. Time-scale analysis approaches are particularly well adapted to represent these physiological components, non-stationary coupled, functioning at different rates and with a spectral hierarchy. Moreover, the treatment proposed for the components differentiation must take into account the intra- and inter-subject variability of the cardiac and ventilatory patterns; to propose a generic solution, we aim at limiting the number of a priori hypotheses. All things considered, we have brought our methodological choices around the empirical mode decomposition (EMD). EMD is a local signal processing technique for representing all the oscillatory modes embedded in a signal without any requirement of stationarity or linearity of the data (Huang et al 1998, Liang et al 2005, Charleston-Villalobos et al 2007). It is an auto-adaptive method, led by the data itself and with minimal parameter definition, which can decompose any complicated signal.
Detecting cardiac blood shift variations from RIP

into a definite number of high-frequency and low-frequency components, called intrinsic mode functions (IMF). The interest of EMD or its variants (Wu and Huang 2009, Yeh et al 2008) has been demonstrated in the CR context with many studies of the literature (Balocchi et al 2004, Bu et al 2007, Abdulhay et al 2009, Fontecave-Jallon et al 2013). Balocchi et al (2004) showed that EMD can be useful for estimating R–R interval variations due to respiration and that these variations are the result of many nonlinearly interacting processes, this highlighting the potential risk of underestimating information content with any linear analysis. In Bu et al (2007), EMD was used to extract local temporal structures such as the heart beats superimposed on respiration signals in order to monitor respiration and cardiac frequencies during sleep using a flexible piezoelectric film sensor. However, it has been shown from experimental results that EMD has one major drawback, known as ‘mode mixing’. In the case of intermittency between the modes involved in a signal, there comes that, after EMD, a same mode is spread over several IMFs or that one IMF contains successive segments of different modes. Variants of EMD, such as the ensemble empirical mode decomposition (EEMD) (Wu and Huang 2009) or complementary ensemble empirical mode decomposition (CEEMD) (Yeh et al 2008), were therefore proposed to overcome this problem, by introducing the Gaussian white noise masking signal. These methods, considered in the CR context, showed that EEMD was a promising nonlinear method for efficient cardiogenic oscillations extraction in simulated CR signals (Abdulhay et al 2009) and that CEEMD provided satisfactory results for the extraction of cardiac volumes from RIP signals (Fontecave-Jallon et al 2013). We will therefore retain the CEEMD method for our current developments on real RIP signals.

In both latter studies, a step of cardiac reconstruction was introduced after the RIP signal decomposition. The physiological attribution of IMFs was mainly made by visual inspection, and the cardiac volume signals were reconstructed by adding the IMFs considered as cardiac. To achieve cardiac information extraction from RIP signals, this step of cardiac volume reconstruction starting from the extracted modes needs to be robust and automatic. It is a crucial and difficult step, since there are no rules established for the choice of the cardiac IMFs. This step will determine whether EMD can be used or not in this context. The optimization and analysis of the cardiac reconstruction will be carried out in section 3, according to physiological concerns about cardiac ejection.

In our current developments, we focus on the volume of blood shift due to heart beat, corresponding to ‘the excess of arterial outflow from the chest over venous inflow during the heart cycle’ as measured by a pneumocardiographic method (Blair and Wedd 1939). This blood shift due to the cardiac activity will be noted $\triangle V_{tr\cdot c}$ in the following and can be assimilated to the cardiac part of the difference between the instantaneous volumes of the body and the trunk, as considered by Aliverti et al (2010) which described the volume of the blood shift as the ‘volume of blood shifted from the splanchnic vascular bed to the extremities by the circulatory function of the diaphragm’.

The cardiac blood shift $\triangle V_{tr\cdot c}$ is estimated from the cardiac volume signal reconstructed from the cardiac IMFs of the CEEMD, and this estimation step corresponds to the third one of our RIP signal processing. $\triangle V_{tr\cdot c}$ is an image of the cardiac stroke volume (SV), which corresponds to the volume of blood ejected by the heart at each cardiac beat; the SV is a major parameter reflecting cardiac activity, since it is the only one that allows the characterization of the cardiac pump and therefore the mechanical monitoring of the heart. The SV should be monitored in widespread applications, such as a long-term monitoring of cardiotoxicity on chemotherapy-treated patients (Sawaya et al 2011), or for longitudinal monitoring of heart rate variability in athletes (Aubert et al 2003).

In such applications, SV measurement methods require that the measurement should be non-invasive to reduce risk and discomfort of the patient, and the sensor should be wearable
to facilitate the device use and ensure better reproducibility. In that context, most gold-standard methods of SV measurements as the thermodilution (Haller et al 1995) or the cardiac catheterization (Gabe et al 1969) are invasive and by consequence not adapted. The state-of-the-art of non-invasive SV measurement methods mainly includes impedance cardiography (IMP) and Doppler ultrasound. This latter measures the beat-to-beat SV, but it has to be operated by a professional practitioner. IMP allows automatic determination of the SV and continuous monitoring (Compton and Schäfer 2009); in addition, it does not require training for its use. IMP measurements were previously validated against gold-standard invasive techniques in many studies (Charloux et al 2000, Moshkovitz et al 2004, Tordi et al 2004, Kemps et al 2008) and are often used for heart activity monitoring.

Therefore, IMP has been chosen for our study as the non-invasive reference SV measurement method and we have already compared SV estimations from IMP and \( \Delta V_{tr,c} \) estimations from RIP in a preliminary work. In Fontecave-Jallon et al (2013), we showed a satisfactory correlation between RIP and IMP for four subjects asked for calm spontaneous respiration. We observed smaller values of \( \Delta V_{tr,c} \) obtained from RIP than SV estimations from IMP, due to the fact that \( \Delta V_{tr,c} \) corresponds to the part of SV ejected out of the trunk. Note also that the SV value estimated from IMP is only indicative and the point is on SV variations in most publications (Panfili et al 2006, Fellahi et al 2009). This suggests that RIP technology should rather be used for cardiac blood shift variations monitoring.

In this study, we therefore analyse the blood volume shift due to heart beat detected by RIP, by comparison with IMP and we propose a specific respiratory protocol assumed to induce significant variations of the SV; this protocol alternates rest respiration and maneuvers adapted from the Vasalva maneuver, as will be described in section 2. Statistical analysis is carried out on volume estimations from RIP and IMP in section 4 and results are discussed in section 5.

2. Material

2.1. Equipment

The equipment configuration diagram is shown in figure 2.

Thorax (THO) and abdomen (ABD) cross sectional area changes were recorded with a computer-assisted RIP vest (Visuresp®, RBI, Meylan, France). RIP relies on the principle that a current applied through a loop of wire generates a magnetic field normal to the orientation of the loop and that a change in the area enclosed by the loop creates an opposing current within the loop directly proportional to the change in the area. The act of breathing changes the cross-sectional area of the patient’s body. With RIP, no electrical current passes through the body. The signal produced is linear and is a fairly accurate representation of the change in the cross-sectional area.

During 3–4 min at the beginning of each recording, breathing was also simultaneously recorded with a flowmeter (Fleish head no.1, Emka Technologies, Paris, France) and a differential transducer (163PC01D36, Micro Switch, Honeywell, USA) placed on a face mask.

Simultaneous measurements were made with a thoracic electrical bioimpedance monitor (PhysioFlow™, Manatec Biomedical, Paris, France). The special instructions for research protocols involving PhysioFlow® were respected. Four impedance electrodes and two electrocardiogram (ECG) electrodes were taped to the skin on the subject’s chest under the RIP vest. The beat-to-beat SV values (SV_{IMP}) were continuously estimated from the impedance signal and an ECG signal was measured (ECG_{IMP}).
Figure 2. The experiment setup. Four impedance electrodes (white circles) and two ECG electrodes (squares) from PhysioFlow system are attached on the chest. Two zigzagging wires (black lines) from the Visuresp system are also placed on the chest. A second ECG acquisition device is placed (black circles) and based on the first derivative of the Lead II ECG signal. Signals are simultaneously recorded, pass through an 1-to-8 converter, and are finally displayed and saved in a desktop PC.

Another ECG signal (ECGREF) was also recorded during the whole recordings (first derivative of the Lead II ECG), and used to check the synchronization of the PhysioFlow and Visuresp systems.

The synchronous acquisition of all signals, sampled at $F_s = 100$ Hz, was realized using a PowerLab 16-bit analogue-to-digital converter and Chart software (A Dinstruments, United States).

2.2. Experiment protocol

As mentioned in the introduction, the aim of this study is to induce variations of the SV and then of the trunk volume due to the cardiac activity.

The changes in posture are ordinarily accompanied by changes in the cardiac output and SV (Naimark and Wasserman 1962). The SV is usually higher in the supine position than in the seated position. The pressure at any point in the vascular system, whether arterial or venous, is affected by the position of the body and the gravitational forces acting on it. When seated man assumes the supine position, blood is transferred from the lower body to the upper body. It follows blood pressure decrease, SV increase and finally cardiac output increase (Coonan and Hope 1983).

Olschewski and Brück (1990) showed that the valsalva maneuver (VM) performed in the supine and seated postures induced a significant SV decrease. Indeed, the VM produces transient but dramatic alterations in venous return, SV, and arterial pressure, due to an increase of the intrathoracic and intraabdominal pressures (Brooker et al 1974). The VM is not easy to perform and can be painful. Then, in order to make manipulations easy to perform, reproducible and quantifiable, we adapted the VM and asked subjects to blow into a U-bend manometer in order to maintain a specific pressure, chosen at 30 cmH$_2$O.

Figure 3 shows a ‘classical’ SV response on the PhysioFlow system while the subject is blowing in a manometer in order to maintain a pressure above 30 cmH$_2$O. At baseline,
the SV is relatively constant (phase 1). With application of expiratory force, pressure rises inside the chest and return of systemic blood to the heart is impeded. The cardiac output is reduced, the SV falls (phase 2) and the heart rate increases. When pressure on the chest is released (phase 3), venous blood can once more enter the chest and the cardiac output begins to increase. The SV usually rises above normal before returning to a normal level. In the same time, the pulse rate returns towards normal (Olschewski and Brück 1990, Brooker et al 1974).

Finally, an experimental protocol (figure 4) was defined in which subjects were asked to alternate spontaneous calm respiration during 2–3 min (noted as rest) and maneuvers consisting in exerting an expiratory effort in a U-bend manometer in order to maintain a pressure higher than 30 cmH2O. One maneuver lasted around 30 s. The protocol was first performed in the seated position and then in the supine position. Each couple rest/maneuver was repeated six times in each position. Participants were asked not to move and not to speak during recording in order to avoid any motion artefacts on signals.

2.3. Subjects

The experiment was conducted in a manipulating room in the TIMC-IMAG Laboratory and this study was approved by the relevant ethics committee (CHU Grenoble). Fifteen healthy volunteers participated in the study after providing informed consent. Their gender, age, mass, height and the hemodynamic parameters measured at baseline, such as systolic blood pressure (SBP) and diastolic blood pressure (DBP) in the seated and supine positions, are summarized in table 1.
Table 1. Physical and hemodynamic parameters of subjects at baseline.

<table>
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<tr>
<th>Parameters</th>
<th>Value (mean)</th>
<th>Standard deviation</th>
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</tr>
<tr>
<td>Gender (M/F)</td>
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<td>Height (cm)</td>
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<tr>
<td>DBP (mmHg) Seated</td>
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<tr>
<td>SBP (mmHg) Supine</td>
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<tr>
<td>DBP (mmHg) Supine</td>
<td>69.2</td>
<td>9.9</td>
</tr>
</tbody>
</table>

2.4. Calibration of IMP and RIP devices

The PhysioFlow system of IMP was used as the reference SV measurement method. The device provides a continuous estimation of the SV ($SV_{IMP}$), which is computed by a specific beat-to-beat calculation method described in detail in Charloux et al (2000). Before the recording of each subject, it was necessary to calibrate the IMP device with physical and hemodynamic parameters at baseline (table 1). The calibration phase was performed for each position (seated and supine) when the subject was completely in the resting state.

As already said, abdomen and thorax cross-sectional area changes and airflow signal were simultaneously recorded at the beginning of each recording. Starting from these signals, the least-squares method of Eberhard et al (2001) was applied to obtain a calibrated signal of thoracic volume variations ($V_{RIP}$). The calibration was performed for each position.

3. Methodology

The principle of the proposed method is to decompose each $V_{RIP}$ signal into different components of different frequencies, reconstruct a cardiac signal from the extracted cardiac components and finally estimate $\Delta Vtr_c$ values. All the processing was computed using MatLab 7.6.0. (MathWorks®, USA).

3.1. Decomposition of the $V_{RIP}$ signal

As justified in introduction, the EMD technique was applied on sequences of $V_{RIP}$ signals. A variant of the EMD algorithm, named CEEMD (Yeh et al 2008) was used.

The principle of EMD (Huang et al 1998) is to decompose a signal into its oscillatory components from the highest embedded frequency to the lowest one. It yields a collection of IMFs characterized by their instantaneous frequency and a residue corresponding to the dc component of the signal. IMFs must verify two properties: (1) zero-local mean, (2) a number of zero-crossings and a number of extrema differing at most of one. The overall signal can be reconstructed by adding up all the IMFs and the residue.

CEEMD is based on the addition of a white noise signal to the original signal $x$ before performing EMD to make the scales uniformly distributed. For CEEMD, each white noise signal is used twice: for a collection $(n_i)_{1 \leq i \leq N}$ of $N$ white noises signals, EMD is applied to each $x+n_i$ and each $x-n_i$. For a given scale, the $N$ resulted IMFs are then averaged to converge towards the true IMF. CEEMD was applied with $N = 100$ white noise signals and a signal-to-noise ratio of 0.6.

For each maneuver repetition, we considered a 20 s stationary sequence at rest just before the maneuver, denoted as the rest sequence, and a 20 s sequence during the respiratory
Figure 5. IMF1-10 (L) obtained after CEEMD applied on one rest sequence of one V_RIP signal. Further IMFs are not shown.

Figure 5 shows the result of CEEMD application to one rest sequence of one V_RIP signal. The ten first IMFs can be observed. In this example, we can consider that IMF1 and IMF2 were
3.2. Cardiac volume signal reconstruction

We focus on cardiac extraction. Therefore, we aimed at extracting a cardiac volume signal from each $V_{RIP}$ signal. According to the physiological attribution of IMFs, a cardiac signal ($V_h$) was reconstructed as the sum of the cardiac IMFs (3–5 in the previous example). This $V_h$ signal can be observed in figure 6, for a part of the rest sequence processed in figure 5.

However, to justify the choice of the cardiac IMFs, a systematic analysis of the cardiac reconstruction had been carried out based on the tracking of possible cardiac ejection, after each R-wave of the ECG signal simultaneously recorded. We considered $E = \{E_j\} = \{\{1\}, \{1,2\}, \{1,2,3\}, \ldots, \{2,3\}, \ldots\}$ the set of possible combinations of successive IMFs and for each 20 s sequence (rest and maneuver for all subjects), we reconstructed all the possible signals $\{V_{h,j}\}, j:

$$V_{h,j} = \sum_{i \in E_j} IMFi.$$

Considering each sequence, the R-waves of the ECG signal were first detected. For each signal $V_{h,j}$ and for each R-wave, we evaluated whether a cardiac ejection could be identified or not on $V_{h,j}$. The criteria of identification of one cardiac ejection were based on the physiological nature of the cardiac ejection process, i.e. a fast (50–200 ms) decrease of the cardiac volume. Therefore, after each R-wave we tried to find on $V_{h,j}$ if there was one segment of 5–20 points ($F_s = 100$ Hz) with a high negative slope. For this purpose, we considered the derivative of the $V_{h,j}$ signal and a threshold equal to the standard deviation of the derivative negative part, as illustrated in figure 6.
Thus, for each $V_{h,j}$, we evaluated the rate $\tau_j$, calculated as the number of ejections identified on the $V_{h,j}$ signal over the number of cardiac cycles detected by the R-waves of the ECG signal.

Results are shown in Figure 7 for the most significant cardiac IMF combinations from $E$. These results correspond to the mean rate of ejection identification over the whole set of 20 s sequences (240 sequences corresponding to 6 sequences of rest and 6 sequences of maneuver per subject in 2 positions). We observe that some IMF combinations provide better ejection identification than others. By fixing a threshold at 75%, a first selection had been made on the combinations of IMFs. A new set of combinations had been limited to $S = \{S_j\} = \{\{3\},\{3,4\},\{3,4,5\},\{4\},\{4,5\}\}$.

Considering each possible signals $\{V_{h,j}\}_j$ from this new set of combinations and the ejections detected according to the ECG signals, we evaluated the volume of blood ejected at each cardiac cycle as the difference between the maximum and minimum of $V_{h,j}$ and averaged these volumes over each 20 s sequence. We observe that this mean value is maximized with the combination of IMF3, IMF4 and IMF5 all sequences considered together, and for 75% of the sequences when considering the sequences one by one, for both maneuver and rest situations.

Therefore and from now, for our cohort of healthy subjects, the cardiac signal extracted from $V_{\text{RIP}}$ and noted $V_h$ was systematically computed as the sum of IMF3, IMF4 and IMF5.

### 3.3. $\Delta V_{tr,c}$ estimation

$\Delta V_{tr,c}$ values were then estimated from $V_h$ signals. Since IMP, which is the reference for our study, requires an ECG signal, the method we employed for cardiac blood shift estimation from RIP also needed the ECG channel reference for the detection of cardiac cycles. Beat-to-beat $\Delta V_{tr,c}$ was estimated by the difference between the maximum and the minimum of each cardiogenic oscillation, detected by the R-waves of ECG (Bloch et al. 1998). Figure 6 shows that RIP seems to be suitable for a beat-to-beat detection of $\Delta V_{tr,c}$; each R-wave of ECG can be associated with one cardiac cycle on $V_h$.

Our aim is to compare $\Delta V_{tr,c}$ estimation from RIP with the SV estimated from IMP ($SV_{\text{IMP}}$). It is worth noting that IMP is rarely used for beat-to-beat SV estimation, but usually averaged on several cycles (Kim et al. 1992, Verschoor et al. 1996). This allows correcting SV variations generated by the respiration. Therefore, 12 consecutive (Guz et al. 1987) $\Delta V_{tr,c}$ (resp. $SV_{\text{IMP}}$) beat-to-beat values were averaged for each sequence of rest and maneuver situations. Over the 20 s of each sequence, we considered for these averages the 12 cardiac beats for which the standard deviations of $\Delta V_{tr,c}$ and $SV_{\text{IMP}}$ were minimized.

The data analysis carried out in the following was realized on these averaged values, from now noted $\Delta V_{tr,c}$ and $SV_{\text{IMP}}$. 

![Figure 7. Mean rate of identifying one ejection at each cardiac cycle on the reconstructed cardiac volume signal $V_h$ for various combinations of successive IMFs and threshold at 75%.](image-url)
3.4. Data analysis

We respectively note $S_R$ and $S_M$ the rest and maneuver situations and $P_1$ and $P_2$ the seated and supine positions. For each subject, we averaged the six $\Delta V_{tr-c}$ (resp. $SV_{IMP}$) values in $P_1$ and the six $\Delta V_{tr-c}$ (resp. $SV_{IMP}$) values in $P_2$ for $S_R$ and $S_M$. Therefore, each subject is associated with eight mean values: $SV_{IMP}(P_1, S_R)$, $SV_{IMP}(P_1, S_M)$, $SV_{IMP}(P_2, S_R)$, $SV_{IMP}(P_2, S_M)$, $\Delta V_{tr-c}(P_1, S_R)$, $\Delta V_{tr-c}(P_1, S_M)$, $\Delta V_{tr-c}(P_2, S_R)$ and $\Delta V_{tr-c}(P_2, S_M)$.

Statistical tests were performed to analyse SV estimations between the various conditions. Since the sample size for each statistical test was lower than $N = 30$, the Wilcoxon signed-rank test, a non-parametric statistical hypothesis test, was used to test the equality of means. A $p$-value $\leq 0.05$ was considered significant for all tests. All statistical analyses were computed using MatLab 7.6.0.

4. Results

Among the 15 recorded subjects, some had difficulties to execute correctly the maneuvers of exerting and expiratory effort into the manometer; this implied a lack of results with IMP technology. Therefore, for the cases corresponding to subject 14 in the seated position $P_1$ and subjects 8, 9 and 15 in the supine position $P_2$, we removed from the dataset the estimations both for IMP and RIP.

4.1. Validation of the proposed protocol according to IMP

In figure 8(a) for the seated position $P_1$ and in figure 8(b) for the supine position $P_2$, averaged $SV_{IMP}$ values for $S_R$ and $S_M$ are displayed for each subject, showing a systematic diminution between rest and maneuver.

The two Wilcoxon signed-rank tests performed to compare $SV_{IMP}$ values between $S_R$ and $S_M$ in both positions indicate that there are significant differences between $S_R$ and $S_M$ both for $P_1$ ($p = 0.02$, $N = 28$) and $P_2$ ($p < 0.01$, $N = 24$). This validates the use of the proposed respiratory maneuver to induce significant variations of the SV. Exerting an expiratory effort during 30 s in a U-bend manometer to maintain a pressure above 30 cmH₂O leads to a decrease of the SV which can be detected by IMP. This SV variation, always in the same direction, is reproducible for every subject.

However, it can be noted that according to two Wilcoxon signed-rank tests, there are no significant differences of $SV_{IMP}$ between $P_1$ and $P_2$ for $S_R$ ($p = 0.47$, $N = 22$) and $S_M$ ($p = 0.79$, $N = 22$).

4.2. Use of RIP for the detection of $\Delta V_{tr-c}$ variations

In figure 9(a) for the seated position $P_1$ and in figure 9(b) for the supine position $P_2$, averaged $\Delta V_{tr-c}$ values for $S_R$ and $S_M$ are displayed for each subject, showing a systematic diminution between rest and maneuver.

The two Wilcoxon signed-rank tests performed to compare $\Delta V_{tr-c}$ values between $S_R$ and $S_M$ in both positions indicate that there are significant differences between $S_R$ and $S_M$ both for $P_1$ ($p < 0.0001$, $N = 28$) and $P_2$ ($p < 0.001$, $N = 24$).

These results highlight that RIP allows the detection of $\Delta V_{tr-c}$ variations, as well as IMP does.

Moreover, it can be noted that according to two Wilcoxon signed-rank tests and similarly to the results with IMP, there are no significant differences of $\Delta V_{tr-c}$ between $P_1$ and $P_2$ for $S_R$ ($p = 0.07$, $N = 22$) and $S_M$ ($p = 0.43$, $N = 22$).
4.3. Correlation between $\Delta V_{tr_c}$ and SVIMP

For correlation analysis between $\Delta V_{tr_c}$ and SVIMP, we consider, for each subject, the averaged values $\Delta V_{tr_c}$ and SVIMP in P1 for SR and SM and the averaged values $\Delta V_{tr_c}$ and SVIMP in P2 for SR and SM. The analysis is therefore carried out on 52 trials.

We observe a correlation ($r = 0.63$, $r^2 = 0.4$, $p < 0.001$) between both methods (figure 10). Cardiac ejection volumes can be discriminated in two groups according to SR and SM. A decrease of cardiac ejection volume is observed for both measurement techniques between rest and maneuver situations.

Moreover, as in Fontecave-Jallon et al. (2013), we observe that $\Delta V_{tr_c}$ values from RIP are lower than the SV estimated from IMP and physiological SV values (Guz et al. 1987).

Finally, correlation results between averaged values of $\Delta V_{tr_c}$ and SVIMP are also presented in table 2, depending on the position (P1 or P2) and the situation (SR or SM). Significant correlations are observed for both situations (rest and maneuver) in the seated position (P1), but not in the supine position (P2).

5. Discussion

Our purpose was to investigate RIP for cardiac functional exploration. RIP technology combined with multi-steps of signal processing enables the extraction of an obviously cardiac...
Figure 9. Diminution of $\Delta V_{tr,c}$ for all subjects between $S_R$ and $S_M$ in positions $P_1$ (a) and $P_2$ (b). (c) Mean ± std of $\Delta V_{tr,c}$ in all conditions.

Figure 10. Correlation plot between averaged values of $\Delta V_{tr,c}$ and $SV_{IMP}$ in both positions ($P_1$ and $P_2$) and for both situations ($S_R$ and $S_M$).
component, since it is consistent with the R-waves of the simultaneous ECG signal (figure 6). Our study focused on cardiac blood shift extraction and showed that $\Delta V_{\text{tr,c}}$ estimated from RIP are in coherence (linear relation, $p < 0.05$) with the SV estimated by IMP both when considering the effect of a maneuver supposed to reduce the SV and when comparing absolute values at rest in the seated position. This position is the one used during classical respiratory function tests.

On average $\Delta V_{\text{tr,c}}$ is lower than the SV, which was expected since $\Delta V_{\text{tr,c}}$ is only an image of the SV and corresponds to the part of the SV ejected out of the trunk. This redistribution phenomenon may be modified in the supine position ending up with the absence of a correlation between the SV estimated by IMP and $\Delta V_{\text{tr,c}}$ estimated from RIP in this position (table 2).

However, we show that RIP can be used to detect the variations of the SV as does impedance technology. RIP can then be considered for cardiac functional exploration via an image of the SV and can provide a non-invasive tool for pooled cardio-circulatory and respiratory function tests. RIP technology thus gives access to both the cardiac and respiratory activities and would allow the study of CR interactions, for example, for patients suffering from sleep apnea syndromes.

Note that IMP estimations of the SV are based on a model which requires physical and hemodynamic parameters at the baseline. IMP, ‘calibrated’ by reference to a cardiac output, is artificially adjusted to the quantity it is supposed to evaluate. In contrast, the physical quantity $\Delta V_{\text{tr,c}}$ measured by our method is a volume calibrated from the pneumotachographic volume reference, independently from any cardiac output measurement.

The results of our study attest the effectiveness of the proposed signal processing method on $V_{\text{RIP}}$.

The method is robust and can be applied to each subject without any specific parameterization. The whole process, including steps of EMD, reconstruction and estimation, can be transposed to new subjects and new experimental protocols. For this study, we have considered healthy volunteers; the algorithm (and in particular the choice of cardiac IMFs) can be directly applied in further studies implying young healthy individuals. We now have to consider situations with patients, for which there might be adjustments of the algorithm. Thus, a next step to be addressed is to evaluate our method in pathological situations, and more precisely to compare, in a clinical study, RIP with the thermodilution method, the invasive gold standard method to estimate the SV.

Currently, the proposed method requires an ECG channel reference for the detection of cardiac cycles necessary for the SV estimation. However, in the purpose of proposing an integrated tool for CR monitoring, this question of cardiac cycle detection without the ECG signal is crucial and will be addressed in future work. Our preliminary investigations indicate that we may do without an ECG signal reference, unlike the concurrent methods such as the ensemble averaging of Sackner et al (1991a, 1991b) or filtering of Bloch et al (1998).

In this study, we have considered $\Delta V_{\text{tr,c}}$ values averaged over 12 cardiac cycles. In future work, it will be interesting to exploit beat-to-beat $\Delta V_{\text{tr,c}}$ estimations, available thanks

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<tr>
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<th>P₁</th>
<th>P₂</th>
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<tbody>
<tr>
<td>SR</td>
<td>$p &lt; 0.05$ &lt;br&gt; ($r = 0.64, r^2 = 0.4, p = 0.01, N = 14$)</td>
<td>n.s. &lt;br&gt; ($r = 0.51, r^2 = 0.26, p = 0.09, N = 12$)</td>
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<tr>
<td>SM</td>
<td>$p &lt; 0.05$ &lt;br&gt; ($r = 0.58, r^2 = 0.34, p = 0.03, N = 14$)</td>
<td>n.s. &lt;br&gt; ($r = 0.09, r^2 = 0.008, p = 0.78, N = 12$)</td>
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**Table 2.** Correlation between averaged values of $\Delta V_{\text{tr,c}}$ and SV IMP depending on position (P₁ and P₂) and situation (SR and SM).
to the method, and to highlight the variations of the cardiac blood shift along the respiratory cycle. Due to the variations of intra-thoracic pressure between inspiration and expiration, the SV decreases during inspiratory phases and increases during expiration, and values could therefore be averaged on specific positions of the respiratory cycle rather than on continuous time periods.

To induce SV variations, we proposed a specific experimental protocol. The SV diminution obtained between rest and maneuver situations was expected and attests the use of the established protocol. The cardiac blood shift variations induced by the protocol are detected with RIP in both the seated and supine positions, where blood circulation does not act similarly, and for each subject. This last point is particularly interesting for a final goal of individual functional tests.

The SV decrease induced by the adapted VM protocol is caused by an increase of pressure inside the subject’s chest, which may cause a stiffening of the rib cage. This may affect the $\Delta V_{tr,c}$ estimation, by decreasing the trunk volume variations due to cardiac activity and thus over-estimating the diminution between rest and maneuver and artificially increasing the sensibility to the protocol; validation studies have to be carried out in various other conditions, namely using cold pressor stress and/or vasopressor agent to avoid the problem of rib cage stiffening. In any case, considering that the shape of the extracted cardiac signal is close to the ventricular volume curves described in breath holding, and this even during rest respiration, and the correlation between $\Delta V_{tr,c}$ and $SV_{IMP}$, we consider that RIP technology can be a detection tool for SV variations in specific protocols.

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