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IFAC PapersOnLine 56-2 (2023) 7359–7364

Wearable Tidal Volume Determination via one Circumferential Measurement and three Strain Gauges – a Pilot Study

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Abstract: The measurement of tidal volumes via respiration-induced upper body surface motions remains a subject of research. Improved and miniaturized sensors allow smart garments to become more common and increasingly used in medical diagnostics and therapy monitoring. Based on the data of a motion capture system, a regression analysis in combination with a bootstrapping technique showed that three circumferential and one distance changes carried the majority of respiratory volume information. These parameters were used for tidal volume calculation, and subsequent evaluation illustrated that tidal volumes can be precisely determined (R^2 = 0.97). For home care monitoring, such a Smart-Shirt would be a valuable alternative to existing devices. However, capturing the various respiratory dysfunctions in a clinical setting may require more sensors.

Keywords: Wearables, Smart-Shirt, Sensor selection, Regression, Lasso operator.

1. INTRODUCTION

Despite recent research efforts, the gold standard measurement of tidal volume is still based on airflow measurement by spirometry (Hayes and Kraman, 2009; Miller et al., 2005) or body plethysmography (Coates et al., 1997; Criée et al., 2011). Hence, for airflow measurement, the use of a face mask is usually required, or the subject must breathe through a mouthpiece while the nose is blocked by a nose clip. This can be uncomfortable especially for long-term measurements. This discomfort can even affect the measurement results themselves (Gilbert et al., 1972).

Alternative tidal volume measurement methods, based on respiration-induced upper body surface motion have been a part of research for decades. While the fundamentals and physiological basis of the respiratory movements of the human upper body were formally presented very early (Sibson, 1848), Wade et al. (WADE, 1954) performed a more sophisticated analysis of the movements and finally, Laufer et al. (Laufer et al., 2023a) provided an overview of respiration induced movement range of various movement parameters and their correlations to respiratory volume. But it was the work of Konno and Mead that established the real beginnings of this area of research in the 1960s (Konno and Mead, 1967). Although the motivation for this research at that time (inaccuracy of existing measurement methods) was different from that of today (comfortable and convenient measurement for the person being examined), the potential and the advantages of the intended method was already evident.

Thus, over the years, many other research approaches and studies followed but unfortunately with mostly limited success; In all these years, only two measurement methods have become common in clinical practice or in home care - the optoelectronic plethysmography (OEP) (Massaroni et al., 2017; Parreira et al., 2012) and the respiratory inductance plethysmography (RIP) (Heyde et al., 2015). Even though, these two methods are only used sporadically these days for different reasons. While the OEP is based on a very expensive, complex and non-portable motion capture system, the reasons for the sporadic use of RIP are rather the inaccurate measurement results.

New sensors or sensor technologies have opened up many new opportunities and applications in this research area (Kaneko and Horie, 2012). In particular, advancing miniaturization improved the implementation of diverse sensors in garments, and their improved accuracy enabled a variety of new applications. Smart-Shirts and other smart garments are gaining popularity and are increasingly used in medical diagnostics and therapy monitoring (Aliverti, 2017).

Available mobile solutions for respiratory monitoring systems are mostly limited to the measurement of respiratory rate, which can be measured with sufficient accuracy for clinical purposes. Unfortunately, in terms of measuring tidal volumes, this accuracy is not yet achieved. Some studies focused on inertial measurement units (IMUs) (Beck et al., 2020; G. Karacocuk et al., 2019; Rahmani et al., 2021) by measuring accelerations and tilt angles. Other studies were based on strain gauges (Chu et al., 2019), or on optical systems, such as the CiMeD belts (Laufer et al., 2021b), measuring circumferential changes on the upper body through optical encoders to determine tidal volumes (Laufer, 2020). Unfortunately, all these approaches have not yet yielded the breakthrough success.

10.1016/j.ifacol.2023.10.351

During the development of a Smart-Shirt, the number of sensors required is an important factor (Laufer et al., 2021a). Minimizing the number of sensors reduces the complexity and cost of the system. Therefore, this study investigated and evaluated whether tidal volumes can be determined using only one circumferential change and three strain gauges. Surface telemetry acquired from a motion capture system (MoCap) was acquired for five subjects. Measurement data were then analysed using the least absolute shrinkage and selection operator (lasso).

2. METHODS

2.1. Data Acquisition

A MoCap (Bonita, VICON, Denver, CO) with nine infrared cameras (VICON Bonita B10, firmware version 404) was used to capture breath-induced movements of the upper body. For this purpose, the subjects wore a tight compression shirt to which 102 reflective motion capture markers were attached (Fig. 1). 48 of the markers were attached ventrally, 18 laterally, and 36 dorsally to the shirt, in 7 different heights / planes (Fig. 1). The different heights were evenly distributed across the torso and ranged from height 1 (at the level of the thoracic vertebra T1) to height 7, which was at the level of the lumbar vertebra L5. However, it must be mentioned that these heights were only approximations that varied depending on the participant's body shape, especially height.

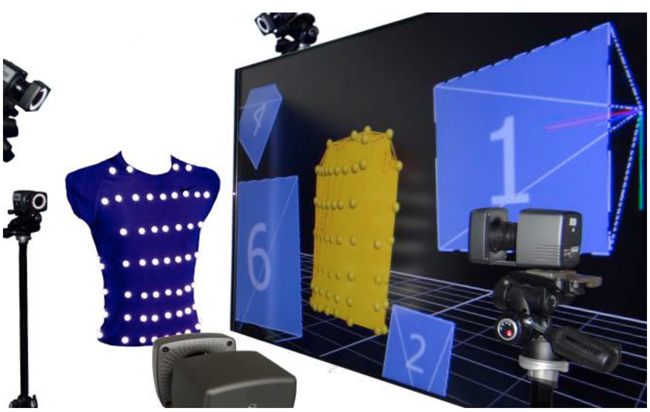

Fig. 1. Sketch of the MoCap system and the used compression shirt (blue).

Parallel to the measurement with the MoCap system, a tidal volume measurement was performed with a spirometer (SpiroScout and LFX Software 1.8, Ganshorn Medizin Electronic GmbH, Niederlauer, Germany) while the subjects performed the required breathing manoeuvres. The raw data from the Mocap system were processed, and the spatial positions of all markers at each time point were transferred from VICON Nexus software (version 1.8.5.6 1009h, Vicon Motion Systems Ltd.) to MATLAB for numerical analysis (R2021a, The MathWorks, Natick, USA).

2.2. Participants and Respiratory Manoeuvres

Five subjects attended the study, three males and two females. Before the measurement, the subjects were fully informed about the study, and informed written consent was obtained from each subject. More details on the subjects are listed in Table 1.

Table 1. Participants

Subject	Height / [m]	Weight / [kg]	BMI / [kg/m^2]	Age / [years]	Gender
1	1.84	75	22.1	18	male
2	1.72	65	22.0	19	female
3	1.70	56	19.4	26	male
4	1.67	57	20.4	18	female
5	1.83	78	23.3	30	male

Similar to Laufer et al. (Laufer et al., 2020), the subjects were asked to perform the breathing patterns shown in Table 2. To reduce movements of the upper body that were not related to breathing, the spirometer was fixed to a rigid mount. This mount was located at the level of the mouth of the subjects sitting upright. Thus, the subjects barely moved their head and upper body during measurement, and the movement data were almost exclusively restricted to respiration-related movements.

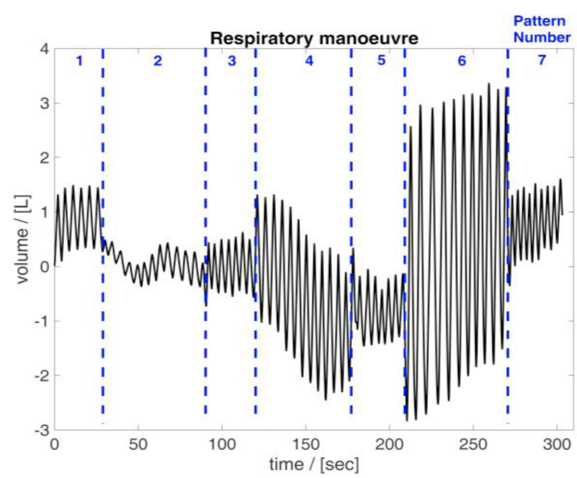

Fig. 2. Volume data obtained by the spirometer during the respiratory manoeuvre. The blue numbers are the breathing pattern identifier according to Table 2. The volume curve was illustrated based on the data of subject 5.

Table 2. Respiratory manoeuvres

Pattern Number	Duration [sec]	Breathing pattern
1	30	spontaneous breathing (normal)
2	60	shallow breathing
3	30	spontaneous breathing (normal)
4	60	medium breaths
5	30	spontaneous breathing (normal)
6	60	maximal breaths
7	30	spontaneous breathing (normal)

At the end of each time in a breathing pattern, subjects were instructed to change the breathing pattern on the next breath. Thus, the exact timing of the measurement depended on the subject and was only an estimate.

2.3. Data Processing

Based on the MoCap data, 361 Euclidean distances between adjacent MoCap markers were calculated and in each of the 7 marker planes the circumferential changes were determined via the length of closed spline curves. The changes in circumferences and distances during the measurement were summarized in matrix form (**A**), and a regression analysis was performed with the aim of determining the reference volume of the spirometer as accurately as possible via these motion data.

Additionally, a bootstrapping resampling procedure was used. For each subject, random parts of the data segment of random length were picked 5,000 times from the measurement data and used for the regression analysis by the least absolute shrinkage and selection operator (Lasso) (Tibshirani, 1996).

The lasso enabled a sparse solution for **x**, which solves

$$\mathbf{Ax} = \boldsymbol{V}_{Spiro}$$

where $\boldsymbol{V}_{Spiro}$ is the reference volume obtained by the spirometer.

The Lasso approach solves the following optimization problem by regularization

$$[x_1, .., x_m]_{opt} = \underset{\boldsymbol{x}}{\operatorname{argmin}} \left(\left\| \mathbf{A}\boldsymbol{x} - \boldsymbol{V}_{spiro} \right\|_2 + \lambda \|\boldsymbol{x}\|_1 \right)$$

where m is the number of chosen parameters and $\lambda\|\boldsymbol{x}\|_1$ is the penalty term of the regularization with λ as regularization factor.

The Lasso operator is generally robust to outliers, delivers sparse solutions and prevents overfitting. The sparse solution results in a minimization of the number of sensors used, which is advantageous, because it reduces the complexity of the Smart-Shirt. A previous study (Laufer et al., 2023b) showed that Lasso is preferable to Ridge regression.

The volume defined by the optimal reduced set of parameters (distance changes or circumferential changes), obtained by the Lasso, was modelled and compared to $\boldsymbol{V}_{Spiro}$. For this purpose, an additional bootstrapping approach was used, which selected 5,000 random segments (of random length) from the MoCap data, modelled the volume $\boldsymbol{V}_{model}$ and compared it with $\boldsymbol{V}_{Spiro}$. Finally, for evaluation purposes, the mean error and mean coefficient of determination R^2 of all bootstrapping steps were determined.

3. RESULTS

Figure 3 shows the sorted rank of the parameters that were most often selected into the set of the best 4 parameters. Data of five subjects and a bootstrapping of 5,000 steps for each subject resulted in a parameter being selected a maximum of 25,000 times into the set of the 4 best parameters. For the purpose of overview, the presentation in Figure 3 has been limited to the 20 most important parameters. The 4 most frequently selected parameters are numbered in Figure 3 and their corresponding locations are shown in Figure 4, where selected distance changes are shown in green and circumferential changes are shown in orange.

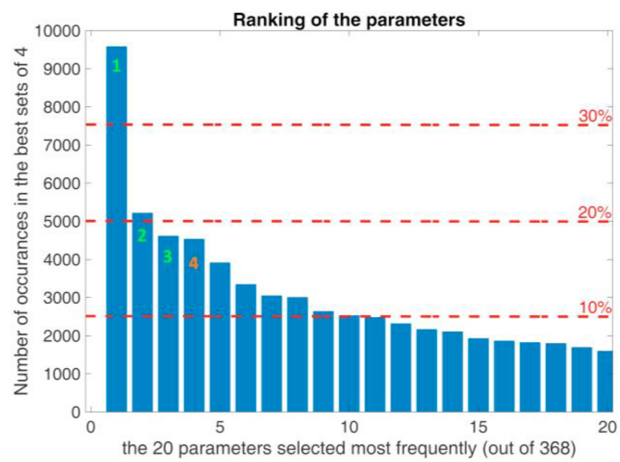

Fig. 3. The sorted ranking of the parameters that were most often selected into the set of 4 best parameters over all 5 subjects (maximal possible: 25,000).

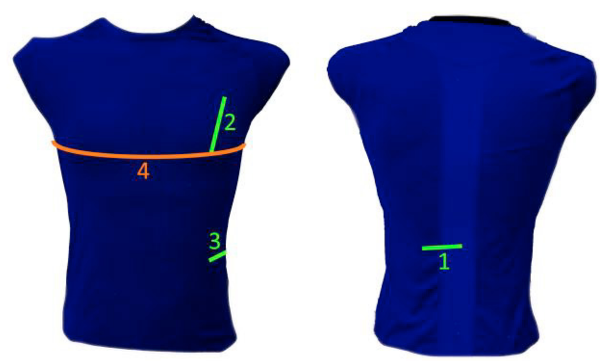

Fig. 4. Illustration of the parameters carrying major respiratory volume information and were selected the most for tidal volume calculation (by Lasso and Bootstrapping). The selected circumferential changes are illustrated in orange, and the selected distance changes are shown in green.

The results of the evaluation are shown in Table 3. Based on R^2 and mean error, the modelled volumes are compared with the reference volume of the spirometer. A random section used for bootstrapping is illustrated in figure 5, where the volume curve $\boldsymbol{V}_{Spiro}$ and $\boldsymbol{V}_{model}$ are shown.

Table 3. Evaluation results of $\boldsymbol{V}_{model}$ compared to $\boldsymbol{V}_{Spiro}$

subject	R^2	Mean error [mL]
1	0.97	115
2	0.97	84
3	0.93	105
4	0.97	60
5	0.96	158

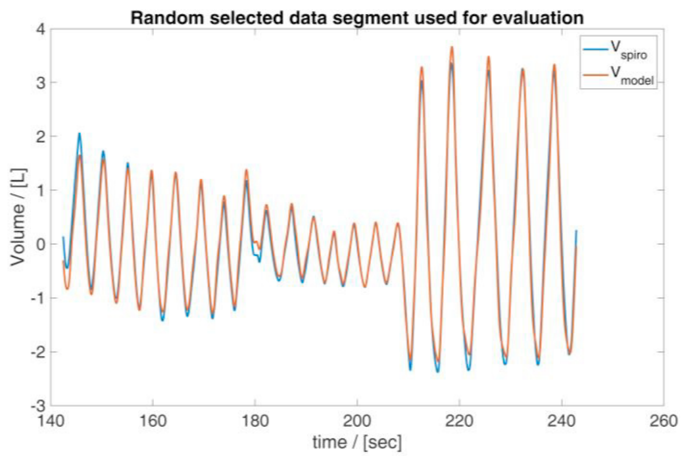

Fig. 5. Example of an evaluation results of $\boldsymbol{V_{model}}$ compared to $\boldsymbol{V_{Spiro}}$, based on the data of subject 5, presented using a randomly selected data segment.

4. DISCUSSION

The use of Smart-Shirts in clinical practice is steadily increasing. In homecare and sports, some vital parameters are already commonly determined via Smart-Shirts. However, the determination of the respiratory volume by means of a Smart-Shirt still shows shortcomings. The Hexoskin Shirt (HEXOSKIN, Montréal, Canada), a Smart-Shirt already available on the market, provides heart rate and respiratory rate reliably, but for tidal volumes, deviations of up to 20% occurred during normal breathing and deviations of more than 40% have been observed during sports activity (Elliot et al., 2019; Villar et al., 2015). Thus, there is still room for improvement.

This study shows a methodology on how to select the optimal number and location of sensors for alternative approaches. The deployment of a lasso in this application area is well suited because the lasso is robust to outliers and prevents overfitting. Moreover, it provides sparse solutions in contrast to other regression / regularization methods (e.g. Ridge regression), which means that Lasso reduces the number of sensors used. Reducing the number of sensors is essential in the development of smart shirts, as this reduces complexity and costs while improving the reliability of the system. However, this is a compromise - an excessive reduction in sensors is usually at the expense of measurement accuracy. The use of the bootstrapping technique additionally reduces the risk of outliers and allows a more robust selection of sensors.

Distances between markers could be measured with textile strain gauges, while circumferential changes require a different measurement method. Laufer et al. developed CiMeD belts (Laufer et al., 2020) for this purpose. Based on optical encoders and code strips, these belts allow accurate measurement of circumferential changes. Thus, based on one of these CiMeD belts and 3 textile strain gauges, the 4 selected sensors would provide a low-cost Smart-Shirt for everyone affordable.

Unfortunately, the result shows that mean errors of up to 160 mL are in a range where they have no clinical relevance. However, as can be seen in Figure 5, these errors are often higher due to different trends of the two volume curves. By focussing optimisation weighting to specifically capture discrete tidal volumes, smaller errors in this metric may be expected. Furthermore, it may be possible to determine certain respiratory parameters, such as the forced expiratory volume in one second (FEV_1), to a higher degree of accuracy than the mean errors of this measurement (Laufer et al., 2022), because some of these values are relative values and independent of the displacement of the curve.

Furthermore, the limitations on the size of the allowable sensor set were arbitrarily chosen. An increase in the number of sensors would trade off cost with improvements in measurement error. Further investigations must show how many sensors would be necessary to ensure the clinical use (error < 5%) of such a smart shirt. However, the presented model with only 4 sensors shows potential to be used for monitoring elderly people in the homecare sector.

The regression performed analyses only linear relationships between the individual motion parameters (changes in distances or circumferences) and the volume of the spirometer. This study was limited to simple linear models as used in regression. More advanced and more complex models integrating non-linear relationships could bring further improvements.

In general, better agreement of the modelled volume with the reference volume of the spirometer was observed for deep and slow breaths. Larger deviations were observed for smaller and faster breaths, thus during shallow breathing, where trend shifts were also observed more frequently. Further analysis of normal spontaneous breathing, which would be the major application especially in homecare, could therefore yield better results and should be investigated in more detail.

Although the results obtained require improvements, the approach used in this study has merit and can be used for further analyses. With this approach, different types of sensors, such as inertial devices (e.g. inertial measuring units IMUs), can be examined for their potential use in determining respiratory volume. Which implies that the number and location of IMUs in a Smart-Shirt for optimal modelling can be determined.

Since breathing is a very individual process, for example, everyone has a different distribution into chest and abdominal breathing, 5 subjects are hardly sufficient for a reliable analysis. Nevertheless, possible trends and improvement possibilities can be identified.

This study was performed based on 5 subjects only. To obtain more reliable results, these measurements should be repeated with more subjects of different sex, age and body shape. Additionally, differences in this respect could be analysed, which could give further and deeper insight in the problem of sensor selection for Smart-Shirts for tidal volume determination.

5. CONCLUSION

The present study shows that a wearable tidal volume measurement via one circumferential measurement and three strain gauges is possible. The results obtained showed errors, which would allow the use of a Smart-Shirt equipped with the four selected sensors for monitoring reasons in the homecare sector. However, for use in clinical practice, further improvements are still required. In particular, the optimum number of sensors for this application should be increased.

ACKNOWLEDGEMENT

This work was partially supported by the German Federal Ministry for Economic Affairs and Climate Action (BMWi) (ZIM-Grant KK5151903BM1) and by the European Commission H2020 MCSA Rise (#872488— DCPM).

AUTHOR'S STATEMENT

Research funding: As mentioned in the Acknowledgments public funded grants are involved. Conflict of interest: Authors state no conflict of interest. Informed consent: Informed consent has been obtained from all individuals included in this study. Ethical approval: The research related to human use complies with all the relevant national regulations, institutional policies and was performed in accordance with the tenets of the Helsinki Declaration and has been approved by the by the Human Ethics Committee of the University of Canterbury (HEC 2019/01/LR-PS) and the Ethikkommission of the Furtwangen University.

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