TITLE PAGE

RUNNING TITLE: Energy expenditure in spinal cord injury

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TITLE

Validation of the use of Actigraph GT3X accelerometers to estimate energy expenditure in full time manual wheel chair users with Spinal Cord Injury.

ABSTRACT

Study Design: cross-sectional validation study.

Objectives: The main goal of this study was to validate the use of accelerometers by means of multiple linear models to estimate the O_2 consumption (VO_2) in paraplegic persons and, secondary, to determine the best placement for accelerometers on the human body.

Setting: Non hospitalized paraplegics' community.

Methods: A volunteer sample of participants (n=20, mean age = 40.03 years, mean weight = 75.8 kg and mean height = 1.76 m) completed a series of sedentary, propulsion and housework activities for 10 min each. A portable gas analyzer was used to record breath-by-breath VO_2. Additionally, four accelerometers (placed on the non-dominant chest, non-dominant waist and both wrists) were used to collect second-by-second acceleration signals. Minute-by-minute VO_2 ($ml\cdot kg^{-1}\cdot min^{-1}$) collected from minute 4 to minute 7 was used as the dependent variable. A total of 36 features extracted from the acceleration signals were used as independent variables. These variables were, for each axis including the resultant vector, the percentiles 10^{th}, 25^{th}, 50^{th}, 75^{th} and 90^{th}; the autocorrelation with lag of 1 second and three variables extracted from wavelet analysis. The independent variables that were determined to be statistically significant using the forward stepwise method were subsequently analyzed using multiple linear models.

Results: The model obtained for the non-dominant wrist accelerations was the most accurate ($VO_2=4.0558-0.0318Y_{25}+0.0107Y_{90}+0.0051Y_{ND2}-0.0061Z_{ND2}+0.0357VR_{50}$) with a correlation coefficient of 0.86 and a root mean square error of 2.23 $ml\cdot kg^{-1}\cdot min^{-1}$.

Conclusions: The use of multiple linear models is appropriate to estimate oxygen consumption by accelerometer data in paraplegic persons. The model obtained to the non-dominant wrist accelerometer data improves the previous published models for this population. In addition, the results show that the best placement for the accelerometer is on the wrists.

Keywords

Paraplegia, physical activity, signal processing, accelerometer, evaluation methodology

INTRODUCTION

People with spinal cord injury (SCI) adopt sedentary habits as a consequence of their disability [1]. Sedentary habits worsen fitness in persons with SCI compared with their able-bodied peers [1] and, in some cases, these individuals present a higher risk of suffering long-term disorders or malfunctions of their organs and systems [2].

Physical Activity (PA) protects against such malfunctions or pathologies [3–5] and is inversely correlated with all-cause mortality. While most of the studies in the literature that have analyzed the relationship between PA and disease prevention have been conducted with able-bodied persons, there are a few epidemiological studies performed in persons with SCI that have shown similar results [6–9]. For this reason, it is very important to know if persons with SCI who perform a minimum level of PA can avoid disorders associated with a sedentary life-style.

To date, most of the studies using able-bodied persons have employed questionnaires to assess PA. This method is inexpensive and easy to administer. Nevertheless, questionnaires present greater subjectivity, and the results depend on the accuracy of the subjects' memories [10].

Other methods employed to estimate PA level based on energy expenditure are indirect calorimetry and heart rate monitors [10]. Due to the high price and the difficulty of employing indirect calorimetry measures in a daily scenario and the low accuracy of heart rate monitoring (during group calibration), neither option is optimal for PA assessment [10].

Another technology used to assess energy expenditure is accelerometry, which is inexpensive, accurate and could be employed in daily activity [10]. In fact, this technique has been one of the most widely accepted method for assessing PA in recent decades and has been validated in numerous recent studies [10,11]. Gracias a estos estudios de validación ya han sido publicados algunos trabajos en los que se ha valorado la actividad física en free-living condition mediante acelerómetros in able-body people [12,13].

In persons with SCI, only a few studies have focused on the relationship between the accelerations and the energy expenditure values [14–18]. Broadly, these studies present some restrictions. For example, the equations were obtained for a restricted number of daily activities, and consequently, the estimation of the energy expenditure in a real scenario could be biased [19]. Likewise, in most of these previous studies, the authors chose integration epochs of 1 minute, which implies having only one feature for the estimation of minute-to-minute energy expenditure.

On the other hand, the placement location is a critical point to estimate energy expenditure from accelerometer. There are studies in persons with disabilities (e.g., multiple sclerosis or chronic obstructive pulmonary disease) that investigate the best placement location [20,21]. This topic should be investigated in spinal cord injured people due to their restricted patterns of movements.

Therefore, the main goal of this study is to validate the use of accelerometers by means of multiple linear models (MLM) to estimate the O_2 consumption (VO_2) in paraplegic persons. Furthermore, this study also aims to determine the best placement of the accelerometer on the human body to obtain the best possible estimation.

MATERIALS AND METHODS

Participants

A consecutive non-randomized sample of twenty subjects whose age, weight and height, in mean (SD), were 40.03 (10.57) years, 75.8 (17.54) kg and 1.76 (0.09) m, respectively, participated in the study. The participants were recruited from the *Hospital la Fe* of Valencia and from the *Asociación Provincial de Lesionados Medulares y Grandes Discapacitados (ASPAYM)*. These subjects were selected using the following inclusion criteria: i. spinal injury between T2 and L5 and diagnosed one year before beginning study participation, ii. full time wheelchair users and iii. completely lost motor ability in their lower extremities (50/100 in ASIA impairment scale). The cause of the injury was traumatic in fifteen of the participants, tumoral in two subjects, iatrogenic in one case, and due to multiple sclerosis and congenital sclerosis in two more cases.

Subjects were excluded if they presented depressive or cognitive disorders; suffered from posttraumatic cervical myelopathy, motor or sensory impairment of the upper extremities, ischemic heart disorder, or recent osteoporotic fractures; had been tracheotomized; or presented sacrotuberous ulcers or hypertension. All subjects gave written consent to participate in the study (approved by the ethical committee of the University of Valencia). We certify that all applicable institutional and governmental regulations concerning the ethical use of human volunteers were followed during the course of this research.

Data collection

All subjects completed a routine of ten activities: lying down, body transfers, moving items, mopping, working on a computer, watching TV, arm-ergometer exercise, passive propulsion, slow propulsion and fast propulsion. These activities of daily living were selected with the objective of having a wide range of intensities of PA and being typical for manual wheelchair users (Table 1). The subjects carried out each activity for 10 minutes with 1-2 minutes of rest between activities. There was only one exception corresponding to the activity of body transfers. In this case, the subjects carried out the activity for one minute and rested for another minute for a total of ten minutes. The transfer task was configured in this way to avoid an overload of the musculoskeletal system in the shoulders.

During each activity, VO_2 was monitored with Cosmed K4b^2 portable (Cosmed, Rome, Italy) gas analysis system. The calibration and placement of the device took into account instructions provided by the manufacturer. This device has been broadly employed as criterion to validate accelerometers. Macfarlane [22] published a manuscript about the validity and reliability of different systems to measure the VO_2 where the readers can check this data for the Cosmed K4b^2. The subjects wore four accelerometers (Actigraph model GT3X, Actigraph, Pensacola, FL, USA): one on each wrist, one on the waist (above the non-dominant anterior superior iliac crest) and the last in the chest (below the non-dominant armpit at the height of the xiphoid apophysis) (Figure 1). The Actigraph was initialized using 1-second epochs, and the time was synchronized with a digital clock so the start time could be synchronized with the gas analyzer.

Signal processing

The Matlab R2010a (Mathworks Inc, Natick, USA) program was used to the preprocessing, segmentation and feature extraction from the signals. The VO_2 signal was preprocessed using averaged blocks of thirty seconds. The time interval between the start of minute 4 and the end of minute 7 was selected. The VO_2 expressed in $ml \cdot kg^{-1} \cdot min^{-1}$ was calculated for each of these minutes. The segmentation of the signals was similar to previous works and confirmed that steady-state VO_2 was reached [23]. The VO_2 for each of the selected minutes was used as the dependent or output variable in the designed models.

The outputs from accelerometers (counts·s^{-1}) were used to obtain predicting variables. Counts are a unit of acceleration used broadly in this topic that represents the amount of acceleration between two consecutive levels of quantization during the analogical-to-digital conversion. We obtained nine total variables for each axis (i.e. X, Y, Z and resultant vector) in minutes number four, five, six and seven of each activity. These variables correspond to features that have been extracted from the time domain and from the Discrete Wavelet Transform (DTW) of the signal. In the time domain, the 10th, 25th, 50th, 75th and 90th percentiles were calculated. Furthermore, as a measurement of the temporal dynamics, the lag-one correlation of each minute was calculated [23].

Finally, three variables were included as a result of the DWT. To present the experimental information in a compressed and arranged format, we have analyzed the signal with multiresolution analysis based on wavelet transform [24,25]. The signal was sampled up to two levels of decomposition using the Daubechies 2 mother wavelet. We calculated the Euclidean norm of the three vectors corresponding to the detail coefficients of the first and second levels of resolution and the approximation coefficients of the second level (i.e. ND_1, ND_2, NA_2). These variables were also included in our analysis (all the descriptive parameters can be seen in the supplementary file).

Mathematical models

We obtained a MLM for each of placement location. We only used statistically significant features determined by the forward stepwise method.

The dependent variable was the consumption of VO_2 in every minute (i.e., 800 values in total). The validation of the model was determined by 20-fold cross-validation. For every model, we computed the following statistical parameters: mean square error (MSE), mean absolute error (MAE), root mean square error (RMSE) and the coefficient of correlation (r). Moreover we calculate the mean error and the percentage error between the estimation and the VO_2 measured by $K4b^2$ for the validation data. Moreover, t-student test for related samples were performed to establish significant differences between criterion and estimate VO_2 values. The level of significance was set at $p=0.05$.

RESULTS

From the analysis of our data, we obtained four linear models with multiple independent variables, one model for each placement location. The models for the waist and the chest have 18 and 11 independent variables, respectively. Due to the large number of independent variables and poor performance of the waist and chest models compared to those corresponding to each wrist, these equations have been included in a supplementary file. Model 1 (equation 1) corresponds to the data obtained from the dominant wrist, while model 2 (equation 2) corresponds to the data obtained from the non-dominant wrist.

$$\begin{aligned} VO_2 &= 4.1355 + 0.0376X_{50} - 0.0155X_{90} - 0.0047X_{NA_1} \\ &+0.0062X_{ND_1} + 0.02Z_{75} - 0.0363Z_{90} + 0.0161VR_{75} + 0.253VR_{90} \end{aligned} \quad \text{Eq. 1}$$

$$\begin{aligned} VO_2 &= 4.0558 - 0.0318Y_{25} + 0.0107Y_{90} \\ &+0.0051Y_{ND_2} - 0.0061Z_{ND_2} + 0.0357VR_{50} \end{aligned} \quad \text{Eq. 2}$$

In these equations, capital letters X, Y and Z represent axes, the sub-indexes represent variables, and VR is the resultant vector. The sub-indexes 25, 50, 75 and 90 are percentiles, and for the variable J, the symbol J_i for i = 25, 50, 75, 90 denotes the value of the i-th percentile of the variable J. The norm of the vector of the approximation coefficients in the first level in DWT is denoted by NA_1, the norm of the vector of the detail coefficients in the first level by ND_1, and the norm of the vector of the detail coefficients in the second level by ND_2. It can be noted that equation 2 has five independent variables, while equation 1 has eight.

The models corresponding to both wrists provide a good estimate of VO_2. The predictions obtained using the accelerometers corresponding to the chest and waist were not very accurate (table 2).

In Figure 2, we show dispersion and Bland-Altman plots for each of the models established. In each case analyzed, no systematic error is observed, but the residuals obtained in the models for the waist and the chest are large (i.e., wider range between ±2 standard deviations).

Additionally, in each of the Bland-Altman plots, there is a tendency to underestimate VO_2 for values larger than 20 $ml \cdot kg^{-1} \cdot min^{-1}$. This tendency is less pronounced for the model corresponding to the non-dominant wrist. Moreover, when we analyzed the activity error expressed as a percent, the relative values obtained were all lower than 20% for the model of the dominant and non-dominant wrist (table 3).

DISCUSSION

The fitting models obtained in the present study improve on the data previously published related to the assessment of PA in paraplegic subjects by means of accelerometry. This improvement can be seen in both the achievement of a stronger correlation between the estimation of VO_2 and the measured value and a lower prediction error for the activities evaluated. We interpret these data to be the result of our use of 1-second epochs for the acquisition of acceleration data.

To the best of our knowledge, there are few studies that have estimated the energy expenditure in persons with paraplegia by means of movement sensors, and most of these studies have used 1-minute epochs of accelerometry data [14–18]. The current study aimed to improve this aspect by including statistical parameters about count distribution during each minute through the acquisition of 1-second epochs. Due to this amount of data (60 per minute), we can perform a feature extraction process and, as a consequence, obtain several variables with relevant information for the estimation of the energetic expenditure.

Moreover, performing 10 different tasks that are representative of daily living provides a wide variety of motion patterns. This variety gives greater consistency to the estimation method obtained. Previous studies only performed sedentary tasks, propulsion by wheelchair and arm-ergometer exercise. Therefore, the estimation methods employed could be insufficient for the assessing of different motion patterns (e.g., transfers, mopping) [14–18].

Of the models generated in our study those of the wrists were more accurate as are expressed by their MAE, MSE, RMES and Pearson coefficients. Moreover the percentage error for each activity was lower for wrists models than for chest and waist equations. This can be due to the reduced mobility of the chest and waist of people with SCI. This fact could uncorrelated the accelerations of these locations with the intensity of the activity.

Regarding the VO_2 values obtained with the gas analyzer from the participants performing the tasks, the data were confirmed to be similar to those provided in previous studies. In the slow propulsion task the consumption measured in our work (i.e., 7.42 $ml \cdot kg^{-1} \cdot min^{-1}$) was almost identical to previous values reported (i.e., 7.35-7.4) when the task was executed at a rate of 4.8-4.9 $km \cdot h^{-1}$ [14,16].

Similar results were also observed in previous studies for other tasks, such as working on a computer [14,26,27], watching TV [27] and moving items [26,27]. Regarding arm-ergometer exercise, we obtained a value of 14.83 $ml \cdot kg^{-1} \cdot min^{-1}$, and we have found values from 7.66 to 20.55 $ml \cdot kg^{-1} \cdot min^{-1}$ in the previous literature, depending on the power developed and the level of the SCI [14,26,27].

The first paper that tried to establish regression equations to estimate the VO_2 in persons with SCI through accelerometry was written by Washburn and Copay in 1999 [16]. They obtained a simple linear equation using the accelerations of the non-dominant wrist with an SEE of 4.99 $ml \cdot kg^{-1} \cdot min^{-1}$. Furthermore, they could explain 44% of the variability of the VO_2 using the $counts \cdot min^{-1}$. Comparing these results with those obtained with the general linear model employed in our study, we can observe some improvements. It is important to note that the estimation errors and r-value depend on the number and type of activities performed to acquire the data use in the validation. Nevertheless to compare between estimators we only have this parameters since are commonly reported in the validation studies. The RMSE in our work is 2.23 $ml \cdot kg^{-1} \cdot min^{-1}$, and the determination coefficient has a value of 0.74. In view of these results work, we found that the use of methodologies that maximize the data available for the estimation of VO_2 can provide general linear models that have better accuracy.

Recently, Hiremath and Ding [17] developed a new equation based on a MLM that was designed using 19 individuals and tested on another 4 for validation. Acceleration data were obtained from the left arm, and indirect calorimetry was employed as a reference measurement during the performance of a limited routine of activities. With the data used to develop the equation (the fitting data set), the authors found an SEE of 1.02 $kcal \cdot min^{-1}$ (2.55 $ml \cdot kg^{-1} \cdot min^{-1}$ approximately) and a r^2 of 0.7. Although these authors improved on preexisting models, the estimation was not as accurate as those models for persons without disabilities. This discrepancy was due to the considerable percentage of error observed for the validation data; this error ranged from 14.12% for arm-ergometer exercise (at 40 W and 90 rpm) up to 113.68% for the resting task.

The MLM of the non-dominant wrist designed in our study have shown values of RMSE and r^2 similar to those obtained in previous studies. However, the percentage of error in each of the activities is lower, and there is less dispersion between activities. Moreover, the minimum and maximum error obtained were 0.67% and 18.68%, respectively. For this reason, the MLM applied in this study provides a methodological improvement for the prediction of the VO_2 in persons with SCI. In our case, the model for persons with paraplegia showed similar estimation errors than the models corresponding to persons without disabilities such as the 2-regression model [28] or ANN based models [23] (although these models were designed with more activities than our model).

The present study does have some limitations. First, although the participants performed a wide range of activities, there are additional activities that should be assessed in future studies (e.g., sports activities as basketball or household activities as washing dishes). Due to the difficulty in recruiting individuals with SCI and the significant administrative burden in the application of all of the protocols, it was not possible to extend the number of tasks executed. In this sense could be interesting to increase also the number of subjects for account with more inter-subjects variability and therefore inprove the robustness of the estimator. Acceleration data have been recorded in counts$\cdot$s^{-1}; raw acceleration data in m$\cdot$s^{-2} would provide more information and therefore a more accurate estimation. Nevertheless we chose 1sec epochs for the memory limitation of the GT3X.

In conclusion, MLM that employ feature extraction from accelerometer signals measured in counts$\cdot$s^{-1} can be used to obtain accurate estimations of the VO_2 in paraplegic persons. Furthermore, it has been demonstrated that in this population, it is possible to record data from either wrist, although there are some benefits of using the non-dominant wrist (i.e., fewer predictive variables and slightly better parameters of performance).

The results of our study could be used to understand PA level in SCI and guide future descriptive studies in this population. The results presented in this work can contribute to identifying patients who are at risk of suffering problems related to a sedentary lifestyle.

ACKNOWLEDGEMENTS

L.M. García-Raffi and E.A. Sánchez-Pérez gratefully acknowledge the support of the Ministerio de Economía y Competitividad under project #MTM2012-36740-c02-02.

X. García-Massó is a Vali+D researcher in training with support from the Generalitat Valenciana.

CONFLICT OF INTEREST.

The authors declare no conflict of interest.

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TABLES

Table 1. Activity routine

Order	Activity	Description
1	Lying down	Lying in the lateral decubitus position
2	Body transfers	Self-shifting the bodyweight from one side to the other using a stretcher (simulating body transfers)
3	Moving items	Loading and transferring boxes with different weights between shelves placed on opposite sides of the laboratory
4	Mopping	Simulation of mopping housework throughout the laboratory
5	Watching TV	Viewing of different television programs
6	Working on computer	Simulation of personal computer work using a word processing program and the internet
7	Arm-ergometry exercise	Performance of an ergometer work sequence with an intensity corresponding to a perception of 8 points based on the OMNI-Res perception scale
8	Passive propulsion	Propulsion of the individual by the researcher
9	Slow propulsion	Self-propulsion of the wheelchair over the floor at a moderate speed
10	Fast propulsion	Fast self-propulsion of the wheelchair over the floor

Table 2. General linear model efficiency of the four accelerometers.

Location	Data	r	MSE	MAE	RMSE
Waist	Fit	0.64	11.33	2.47	3.32
	Validation	0.67	10.61	2.39	3.26
	All	0.67	10.65	2.39	3.26
Chest	Fit	0.66	10.80	2.45	3.26
	Validation	0.68	10.41	2.41	3.23
	All	0.68	10.43	2.41	3.23
Dominant wrist	Fit	0.85	5.32	1.69	2.28
	Validation	0.86	5.16	1.67	2.27
	All	0.86	5.16	1.67	2.27
Non-dominant wrist	Fit	0.86	5.08	1.66	2.23
	Validation	0.86	4.98	1.65	2.23
	All	0.86	4.98	1.65	2.23

r=coefficient of correlation, MSE=mean square error, MAE=mean absolute error, RMSE=root mean square error. Fit corresponds with the data set used to adjust the model. Validation corresponds with the data set used to validate the model. All corresponds with fit and validation data sets together.