



# Monitoring respiratory rates with a wearable system using a stretchable strain sensor during moderate exercise

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## Abstract

Respiratory rate, a sensitive indicator of respiratory status, is rarely measured during the field walking test. Our objective was to develop and validate a non-invasive, wearable monitoring system using stretchable strain sensors and an accompanying algorithm capable of providing real-time measurements of respiration during exercise. Twenty-four healthy volunteers wore stretchable sensors during a walking test protocol that included standing, sitting, walking, and walking with a stick. Sensors were placed on the ribcage and abdomen. The Bland-Altman method was used to assess the accuracy and precision of breath counts; total respiration time and inspiration time ratio were determined by custom algorithms and compared with measurements obtained with the standard flow sensor. The output signal from the stretchable sensor was highly synchronized with flow signals. The limits of agreement were within 3 breaths/min throughout the test protocol. Differences between sensors for total respiration time and inspiration time ratio were less than 14% and 26%, respectively. The agreement was maintained regardless of respiratory rate or volume. The wearable respiratory monitoring system yielded accurate and precise breath counts and total duration of respiratory cycle during moderate exercise in healthy young individuals, suggesting that it might be useful in clinical practice.

**Keywords** Rehabilitation · Stretchable strain sensor · Respiratory rate · Vital signs · Walking test · Wearable sensor · Respiration

## Abbreviations

ABD	Abdomen
BC	Breath count
FLOW	Flow sensor
LOA	Limit of agreement

RC	Ribcage
RMSRD	Root mean square relative difference
STR	Stretchable strain sensor
STS	Sit-to-stand/stand-to-sit
TT	Total time of respiratory duration
TITT	Time of inspiration per total time of respiratory duration (inspiration time ratio)

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## 1 Introduction

Chronic respiratory diseases, particularly chronic obstructive pulmonary disease, have been a major public health problem and remain a challenge in the 21st century [25]. Field walking tests including the self-paced 6-min walking test or shuttle walking tests for patients with chronic respiratory diseases are low-cost, easy to perform, good reflective of daily living, thus suitable for evaluation in community settings as well as hospitals [2]. Recent available guideline admitted the usefulness and reliability and suggested continuous monitoring for vital signs including oxygen saturation and heart rates even during field

walking tests [17, 33]. Although respiratory rate, which is time domain information of respiration, is a good indicator of physical effort during exercise, measuring respiratory rate is uncommon practice in the field walking tests. This is partly because its measurement is challenging especially during exercise.

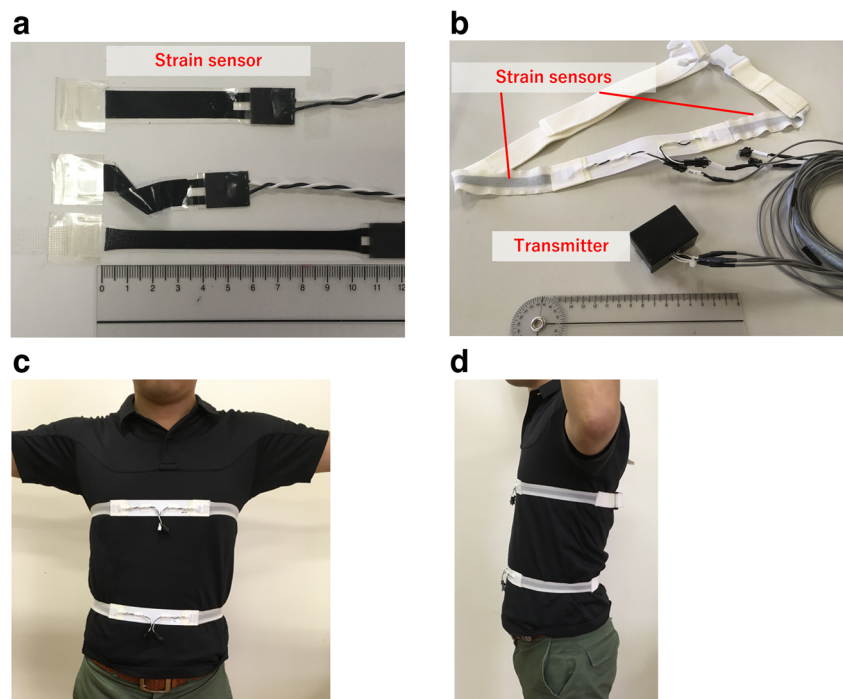
Respiratory rate during exercise is usually measured with a respiratory flow sensor attached to the face mask [9, 28]. However, medical flow sensors are large and expensive, and facial masks or mouthpieces may cause feelings of oppression or influence breathing patterns even at mild workloads [5]. Also, air flow measurements may be disturbed in patients undergoing portable oxygen therapy for severe desaturation. By contrast, wearable sensors attached on the clothing or even mounted on the human skin is expected to be applied for real-time monitoring of human movements, including respiratory rate, without facial contact [1].

Traditionally, chest wall movements are used to monitor breathing without facial contact using various technologies [32]. For instance, transthoracic impedance, respiratory inductive plethysmography (RIP), and flexible strain sensors have been used in both laboratory and clinical settings [8, 14, 16, 23, 24, 26, 31, 34]. For skin-mountable or wearable use, sensors should fulfill several requirements including light weight, flexibility and stretchability, durability, repeatability, and robust to human motion artifacts. Thoracic impedance may provide inaccurate measurements at high skin-electrode impedances [13]; thus, skin contact electrodes require time to

decrease skin electrical resistance and a technique to fix the electrodes on the skin in exercise. RIP requires covering whole circumference of the patient's chest and abdomen using two bands, resulting in cumbersome and less portable.

Recently, small, light and flexible, and stretchable strain sensors have gained more attention for human motion monitoring. Previously, flexible strain sensors classified into resistive-type demonstrated good correlation to respiratory volume curve and feasible to measure respiratory rate in resting state [8, 30]. However, error analysis and data of the agreement to golden standard in time domain information especially during exercise with a large sample size was scarcely available. During exercise, motion artifacts like heel strike and arm swing cause noise and disturb accurate continuous respiratory monitoring. In addition, a monitoring system is required to catch up exercise-induced increased respiratory rate. Also, as far as authors know, an algorithm for real-time detection of respiratory cycle was not developed. Respiratory cycles were detected using digitally smoothed signals which large time delay is inevitable [8]. To overcome the exercise artifacts, we have created the chest-band like sensor system including two capacitive-type strain sensors and developed rapid algorithm for real-time calculation of respiratory cycles (Fig. 1a, b). Then, we set an exercise test protocol involving those motion artifacts and investigated the validity of the novel system as a respiratory monitor to establish the basis for further clinical application studies.

**Fig. 1** Pictures of stretchable strain sensors. **a** Raw stretchable strain sensors whose original length was 10 mm height and 50 mm width. **b** A respiratory monitoring system composed of a band with two sheets of strain sensors, 4 channel of input cables, and a transmitter. Stretchable strain sensors placed around the ribcage and abdomen. **c** Frontal plane. **d** sagittal plane



## 2 Methods

### 2.1 Study protocol

#### 2.1.1 Participants

Twenty-four young healthy individuals (12 males and 12 females) participated in the study. Before the treadmill procedure, all participants completed a questionnaire about their own health status as well as demographic characters. Only subjects without chronic respiratory or cardiovascular disease and orthopedic diseases that might have interfered with the walking test were included. Anthropometrical data were collected, including body height, weight, body mass index (BMI), sex, age, and chest expansion at the level of the xiphoid process and the navel using a tape measure. Informed consent was obtained from all participants prior to the study. This study was approved by the Ethics Committee of the Kobe University Graduate School of Health Science in accordance with the Helsinki declaration.

#### 2.1.2 Experimental procedure

Participants walked on a treadmill in the laboratory. There were six protocol phases. Phase 1: upright stance (3 min); phase 2: upright sitting (3 min); phase 3: normal walking (6 min); phase 4: upright sitting (3 min); phase 5: stick walking (6 min); and phase 6: upright stance (3 min). In normal walking, the treadmill belt speed was changed in the following sequence: 2 km/h (1 min); 4 km/h (1 min); 6 km/h (2 min); 4 km/h (1 min); and 2 km/h (1 min). In stick walking, the belt speed was changed in the following sequence: 2 km/h (1.5 min); 4 km/h (3 min); and 2 km/h (1.5 min). Belt speed was changed without stopping the belt. Since stand-to-sit occurred from phases 1 to 2 and sit-to-stand from phases 2 to 3, the sit-to-stand/stand-to-sit phase was defined as the time between phases 1, 2, and 3. Thereafter, phases were combined to define

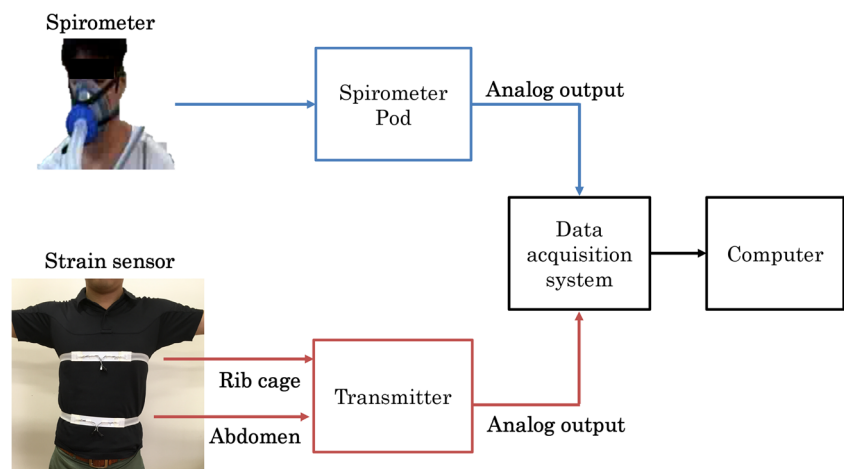
five movement conditions: standing, sitting, walking, stick walking, and sit-to-stand/stand-to-sit for analysis.

#### 2.1.3 Apparatus

We used a stretchable strain sensor to detect thoracic and abdominal expansion associated with respiration (STR; Bando Chemical Industries, Ltd., C-STRETCH®, Hyogo, Japan) [27]. The sensor system was composed of sensors, a transmitter, and input and output cables (Fig. 1b). The detection area of the sensor was 10 mm wide  $\times$  100 mm long, which can extend to almost double in its size. Each sensor is composed of three elastomer layers and two electrode layers, where the parallel plate structure theoretically works as a capacitor. The capacitance of the sensor is linearly related to the strain of sensing area because the elastomer sheets were hardly compressed. The strain of the sensor due to chest expansion with breathing was measured by the change of a chest band length composed of two sensors connected with non-stretchable clothing. The participants wore the band on the sport wear at the level of the 4th rib level (upper ribcage, RC), and the level of the 10th rib (upper abdomen, ABD) (Fig. 1c, d). Each RC and ABD band contained two sheets of sensors placed in left and right side series symmetrically.

A spirometer (ADInstruments, Spirometer Pod, Dunedin, New Zealand) attached to the participant's face mask was used to measure the analog signal of respiratory flow (FLOW) and volume. Volume signal was obtained by numerical integration with zero offset at the start of every respiratory cycle. Expiratory flow was set at positive and inspiration was negative. The strain sensor and spirometer were connected to 16-bit A/D converter (ADInstruments, PowerLab 16/35, Dunedin, New Zealand) for sampling at 100 Hz. Data were stored on a computer for off-line analysis. The experiment was simultaneously recorded using a video camera. The experimental setup was summarized in Fig. 2.

**Fig. 2** Schematic of the experimental hardware setup for human subject test. The spirometer was attached on the mouth with a facial mask. The strain sensors implemented in a band were placed on ribcage and abdomen at the level of 4th and 10th ribs respectively. Airflow was measured by the spirometer pod and processed to a data acquisition system. Through the transmitter circle, capacitance of the strain sensor was outputted to the data acquisition system



## 2.2 Data analysis

### 2.2.1 Detecting the respiratory cycle

For the FLOW signal, the onset of each respiratory cycle was determined using measurement software (Spirometry extension module, LabChart 8.0, and Spirometry module, ADInstruments, Dunedin, New Zealand). In each respiratory cycle, FLOW data were numerically integrated, and the onset of expiration was identified by a local maximum point of the integrated data. The identified respiratory cycle and onset of expiration were confirmed using visual inspection of FLOW signal and video recordings. Coughing, speaking, or motion artifacts that were mistakenly recognized as respiratory cycles by the system were manually excluded. We divided the respiratory cycle into two phases: namely inspiration and expiration.

For STR, we created a time domain algorithm to discriminate between inspiration and expiration from the signal. Figure 3 shows the flow chart of the algorithm. To achieve real-time analysis, we did not use pre-processing of the signal or smoothing.

Step 1. Data of two sensors in each RC and ABD were combined to increase signal/noise ratio.

Step 2. Based on the current phase, inspiration or expiration mode was selected.

Step 3. The differences between current and previous data were obtained to determine the upward or downward trend of the signal as follows:

$$dx[t] = x[t] - x[t-1]$$

Step 4. During expiration, if the current different value was positive ( $dx > 0$ ), the system start searching local minima. If there were not smaller values than the current values within the predefined time constant, the current value was considered as a candidate of the local minima.

Step 5. This step was to confirm the decision above. If the current value was smaller than the threshold ( $p1$ ), the value was finally defined as local minima and breath phase was changed to inspiration.

Step 6. Only when the phase changed from expiration to inspiration, the threshold and the time constant were updated based on a mean amplitude and mean duration of previous respiration cycles.

In the phase of inspiration, step 4 and step 5 are processed for searching local maxima as a time point of phase shift.

### 2.2.2 Outcomes

The primary outcome was the number of breaths counted in every minute (breath counts). Detection of the respiratory cycle and phase was the secondary outcome. For the secondary outcome, the moving average of the total duration of the respiration cycle (TT) was computed breath by breath using the previous eight cycles. In addition, to examine the validity of respiratory phase detection, the moving average of the inspiration time ratio (TITT) was calculated, breath by breath, using the previous eight cycles:

$$TITT = \frac{1}{8} \sum_{i=1}^8 \frac{T_{IN,i}}{T_i}$$

where  $T_{IN}$  represents the duration of inspiration, and  $TT$  represents total duration of the respiration cycle.

### 2.2.3 Statistical analysis

The data are expressed as mean  $\pm$  SD. Agreement of breath parameters between FLOW and STR were evaluated using Pearson's product-moment correlation coefficients and Bland-Altman plots and their analysis for repeated-measures samples [3]. A Bland-Altman plot is formed by plotting the differences ( $d$ ) on the vertical axis and the average (STR – FLOW)/2 on the horizontal axis. The time series of differences between outcomes, namely breath count, TT, and TITT, from STR and FLOW were obtained with the following equation.

$$d(t) = STR(t) - FLOW(t)$$

In addition, relative differences between sensors (RD) for breath count, TT, and TITT were quantified using root mean square relative difference (RMSRD) as follows.

$$RD(\%) = \frac{STR - FLOW}{FLOW} \times 100$$

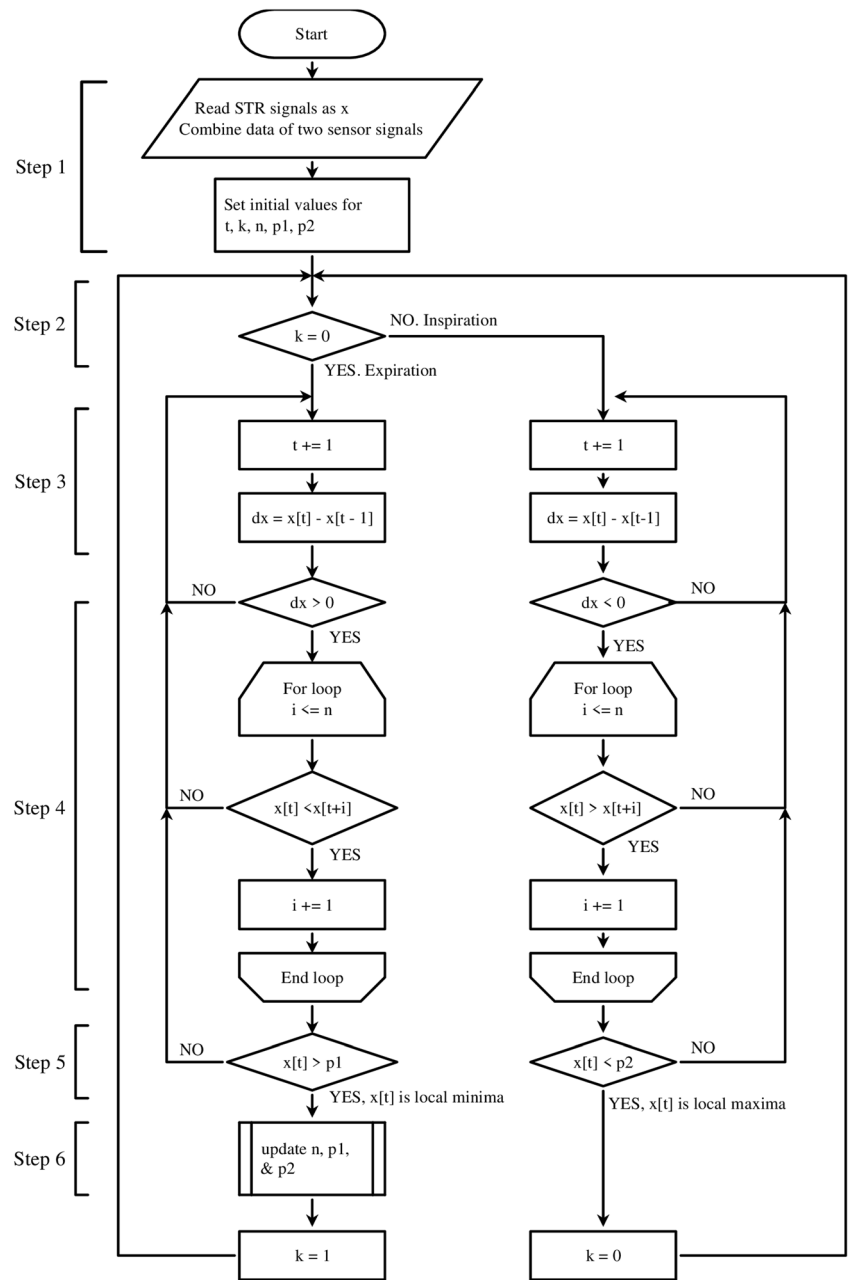
$$RMSRD(\%) = \sqrt{\frac{1}{n} \sum RD^2}$$

For the Bland-Altman plot, three horizontal lines were drawn, one for the bias ( $\bar{d}$ ), which is computed by the mean of the difference  $d$ , and the other two for upper and lower limits of agreement (LOA), which were calculated as follows:

$$LOA = \bar{d} \pm 1.96s_d$$

Considering inter-individual difference in breath counts, standard deviation ( $s$ ) to obtain LOA was calculated using the multiple observation technique suggested by Bland and Altman [4]. Based on a one-way analysis

**Fig. 3** Flowchart describing the algorithm developed to detect breathing cycles



of variance, we have  $m$  pairs of observations for subject  $i$ , and there are  $k$  subjects. The estimated variance of the difference for pairs of measurements ( $\theta_{d}^2$ ) was provided by the sum of within-subject mean square ( $MS_W = \theta_{dW}^2$ ) and the variance for the differences between the average difference across subjects ( $\theta_{dI}^2$ ).

$$\theta_d^2 = \theta_{dI}^2 + \theta_{dW}^2$$

$$\theta_{dI}^2 = (MS_b - MS_W) / \frac{(\sum m_i)^2 + \sum m_i^2}{(k-1)\sum m_i}$$

where  $MS_b$  is the between-subject mean square and

$MS_w$  is the within-subject mean square. The square root of the total variance was used for calculating LOA.

For both bias and LOAs, their 95% confidential intervals (CI) were computed and compared to limits of equivalence (LOE) in all data as well as in each movement condition (standing, sitting, walking, stick walking, and sit-to-stand/stand-to-sit). According to confidence interval approach of equivalence, equivalence is assumed if the confidence interval of an observed difference lies entirely within a specified equivalence range [19], which was defined as  $\pm 3$  counts per minute for breath counts. The LOE for breath counts was determined based on the accuracy information of the existing

respiratory monitors [21, 29], and the opinions of pulmonary rehabilitation specialists. LOE for RD of breath count, TT, and TITT was set at 10% according to the criteria of respiratory inductive plethysmography [7, 14].

The effect of respiratory cycle duration and tidal volume to the difference between STR and FLOW measurements was analyzed by using Pearson's correlation analysis.

The effects of BMI and chest expansion on the RMSRDs was determined using Spearman's rank-correlation analysis. The difference of RMSRDs between male and female was examined using the Wilcoxon rank sum test. The statistical significance level was set to  $P < 0.05$ . The data were analyzed using MATLAB® R2017b (MathWorks, USA) and R ver. 3.31.

## 3 Results

### 3.1 Subjects and breath performance data

Characteristics of the 24 participants are described in Table 1. Chest expansions averaged 5.6 cm at the xiphoidal process and 4.7 cm at the navel.

Figure 3 shows typical recordings from a female participant during standing to sitting (Fig. 4a) and walking with different belt speeds (Fig. 4b). The STR signal highly synchronized with the FLOW signal with a constant phase lag. The STR signal increased during inspiration and decreased during expiration, while FLOW decreased during inspiration and increased during expiration. When participants moved from standing to sitting the baseline level of ABD signals increased due to the increase in abdominal circumference. As walking speed increased from 4 to 6 km/h, both STR and FLOW measured shortened respiratory cycles. Looking at each local maxima and minima of STR signals during walking, the

signal was noisier compared to that during standing or sitting. The peak of each respiratory cycle was ragged and less clear. The detected respiratory phases during walking (Fig. 5a) and during sitting to standing (Fig. 5b) were illustrated with a typical waveform.

Table 2 summarizes breath performances in the five movement conditions. When compared to other conditions, breath counts were higher and TT was shorter in walking and stick walking. In the sit-to-stand/stand-to-sit condition, a slight increase in breath counts and decrease in TT was observed relative to standing and sitting values. Inspiratory time ratio was slightly increased during normal walking (mean 49.5%).

### 3.2 Accuracy in counting breath

We compared breath counts between the FLOW and STR signals for the five movement conditions (Table 3). Total numbers of breaths measured by FLOW, RC, and ABD were 13,414, 13,444, and 13,488 respectively. The overall performance of the breath counting system in RC was 0.0 counts/min, LOA [− 1.9 to 1.9] and for ABD 0.1 counts/min LOA [− 1.6 to 1.8]. There was a strong correlation among the values for breath counts per minute with  $r$  exceeding 0.99 (RC,  $r = 0.991$ ,  $P < 0.001$ ; ABD,  $r = 0.993$ ,  $P < 0.001$ ). Of all computed periods, more than 98% in RC and 99% in ABD were within  $\pm 3$  counts/min difference of LOE. Biases and 95% confidential intervals were  $< \pm 0.6$  counts/min for all movement conditions, indicating no additional error. Limits of agreement were  $< \pm 3$  counts/min for all conditions and for all sensor parts (LOAs of RC ranged from − 2.0 to 2.1 in RC and from − 1.8 to 2.0 in ABD) (Fig. 6a, d).

### 3.3 Accuracy in measuring respiration cycle and inspiration time ratio

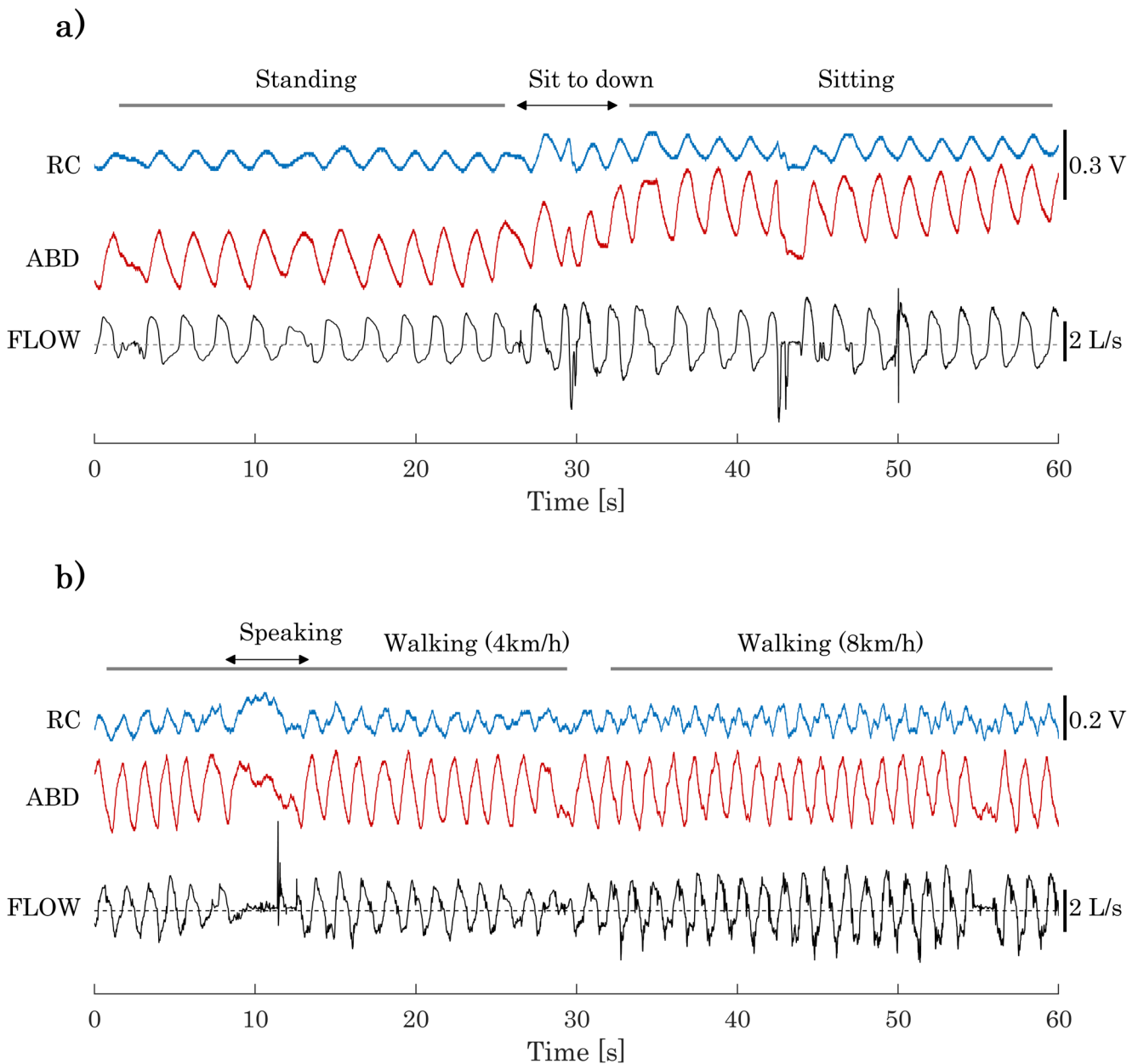
Total duration of respiratory time was compared between FLOW and STR (Table 4). The overall detection performance was 0.63%, 95% LOA [− 12.3 to 13.6], in RC and 0.16%, 95% LOA [− 10.2 to 10.5] in ABD. There was no bias between FLOW and STRs in both RC and ABD during all movement conditions as 95% CIs included zero. Each pair of TTs were very strongly correlated between FLOW and STR (for RC,  $r = 0.98$ ; for ABD,  $r = 0.99$ ). For RC and ABD, respectively, for all test periods, 91% and 93% of all breaths were accurately detected within the predefined LOE, 10% of difference in TT (Fig. 6b, e).

Phase detection accuracy was investigated using TITT values. Overall performance of TITT detection was follows; bias [95%LOA] was − 1.9 [− 19.6 15.7] and 8.8 [− 25.2 7.6] (%) in RC and ABD of STRs, respectively (Fig. 5c, f and Table 5). A linear relationship was weak between TITT ratios of STR and FLOW for both RC ( $r = 0.17$ ) and ABD ( $r =$

**Table 1** Descriptive characteristics of all participants

No. participants		[n]	24	
	Female	[n]	12	
Age*		[year]	24	(6.1)
Height*		[m]	1.64	(0.1)
Weight*		[kg]	57.5	(10.6)
BMI*		[kg/m <sup>2</sup> ]	21.3	(3.2)
Circumstance at				
Xiphoid process*	Inspiration	[cm]	88	(5.8)
	Expiration	[cm]	82.4	(6.1)
	Expansion	[cm]	5.5	(2.0)
Navel*	Inspiration	[cm]	73.9	(7.7)
	Expiration	[cm]	69.2	(8.0)
	Expansion	[cm]	4.6	(2.2)

\*Data are mean and SD. BMI body mass index



**Fig. 4** Typical waveforms of respiratory signals obtained by stretch and flow sensors. **a** standing to sitting, and **b** walking at 4 to 6 km/h. Top black lines are signals from stretch sensors around the ribcage; middle gray lines are signals from stretch sensors around the abdomen; bottom

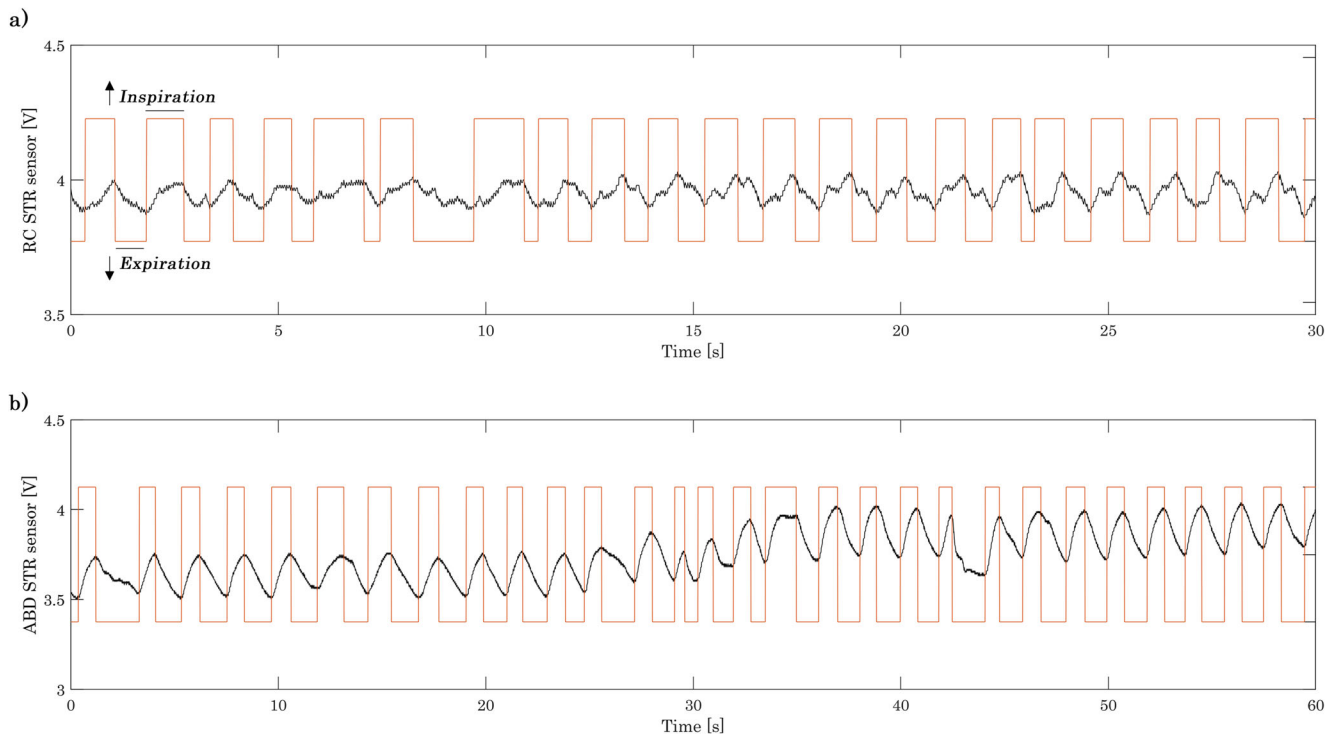
lines are flow sensor signals. RC and ABD are signals of stretch sensor attached at the ribcage and abdomen, respectively. FLOW is the signal obtained by the flow sensor

0.24). Among all breaths, 67% (RC) and 62% (ABD) were within the 10% difference LOE. Considering 20% of difference from FLOW, 91% (RC) and 88% (ABD) of TITT values were within the range.

### 3.4 Error analysis in comparison with respiratory cycle and volume

The influence of respiratory cycle and volume to the accuracy measurement was investigated and shown in Figs. 7 and 8.

Figure 7 shows the difference between STR and FLOW had no linear relationship with tidal volume in all breath counts, TT, and inspiratory ratios. In fact, correlation coefficient values were all within  $|r| < 0.2$ . When examining the relationship between the difference measurements and respiratory cycles, there was no clear relation between them (Fig. 8). Most of difference measurements were distributed around zero regardless of respiratory speed. This observation was supported by the correlation values which were less than  $|r| < 0.2$ .



**Fig. 5** Typical waveforms of respiratory signals and detected phases obtained by stretchable strain sensors. **a** Signals from stretch sensors around the ribcage during walking at 4 to 6 km/h. **b** Signals from sensor around abdomen during standing to sitting. Red lines indicate

detected phases. The periods which the phase signal is upper (lower) area are detected inspiration (expiration) phases. The waveform was from the same subject in Fig. 3

### 3.5 Physical and sexual effects to accuracy of the measurements

The effects of physical and sexual factors on the agreement between STR and FLOW were investigated (Table 6). There was no significant correlation between the RMSRD of breath counts and chest expansion and BMI in both RC and ABD ( $\rho = -0.36\text{--}0.27$ ; all  $P > 0.05$ ). Two sample  $t$  tests revealed that the mean difference in RMSRDs of breath counts between male and females was not statistically significant (all  $P > 0.05$ ).

Similarly, there was no significant effect of physical and sexual factors on the detection performance of respiratory

cycles and phases. The RMSRDs of TT and TITT were correlated neither with chest expansion nor BMI ( $\rho = -0.07$  to  $0.21$ , all  $P > 0.05$ ). There was no significant difference in RMSRDs of TT and TITT between males and females for both RC and ABD (all  $P > 0.05$ ).

## 4 Discussion

We evaluated the validity of the novel stretchable strain sensors to detect respiration during treadmill walking. High accuracy of STR-derived breath count estimates and TT was

**Table 2** Descriptive statistics of breathing in the five conditions

		Standing	Sitting	Walking	Stick walking	STS
Breath count	Breaths [count]	2547	1771	3576	3960	1560
	Mean [/min]	18.2	18.8	28.8	27.9	21.7
	SD [/min]	5.2	4.8	6.8	5.2	5.2
TT	Mean [s]	3.3	3.2	2.1	2.1	2.8
	SD [s]	1	0.8	0.5	0.5	0.8
TITT	Breaths [count]	2546	1771	3568	3960	1560
	Mean [%]	45.5	46.5	49.5	46.3	46.6
	SD [%]	5.5	5.1	4.5	2.9	5.2

Data were obtained from flowmeter. *STS* sit-to-stand/stand-to-sit, *TT* total duration of the respiration cycle, *TITT* inspiration time ratio, *SD* standard deviation



**Table 3** Breath count comparison between flow and stretchable strain sensors by conditions

Sensing part	Condition	n	Breaths Counts	Bias Counts/min	SD Counts/min	Lower LOA (counts/min)		Upper LOA (counts/min)	
						Mean	95% CI	Mean	95% CI
Ribcage	Standing	24	2536	0.1	1.1	-2.0	[-2.3, -1.7]	2.2	[1.9, 2.5]
	Sitting	24	1761	-0.1	0.8	-1.7	[-2.0, -1.5]	1.5	[1.4, 1.8]
	Walking	24	3594	0.1	1.0	-1.8	[-2.1, -1.6]	2.0	[1.8, 2.3]
	Stick Walking	24	3967	0.2	1.1	-1.9	[-2.2, -1.7]	2.3	[2.1, 2.6]
	STS	24	1538	-0.3	0.8	-1.9	[-2.2, -1.7]	1.3	[1.1, 1.6]
Abdomen	Standing	24	2543	0.1	0.9	-1.8	[-2.0, -1.6]	2.0	[1.8, 2.2]
	Sitting	24	1760	-0.1	0.8	-1.7	[-1.9, -1.5]	1.5	[1.3, 1.7]
	Walking	24	3606	0.2	0.9	-1.5	[-1.8, -1.3]	1.9	[1.7, 2.2]
	Stick walking	24	3971	0.3	0.8	-1.3	[-1.5, -1.1]	1.9	[1.7, 2.1]
	STS	24	1560	0.0	0.8	-1.6	[-1.9, -1.4]	1.6	[1.4, 1.9]
Overall	Ribcage	24	13396	0.0	0.9	-1.9	[-2.1, -1.6]	1.9	[1.6, 2.1]
	Abdomen	24	13440	0.1	0.9	-1.6	[-1.8, -1.4]	1.8	[1.6, 2.0]

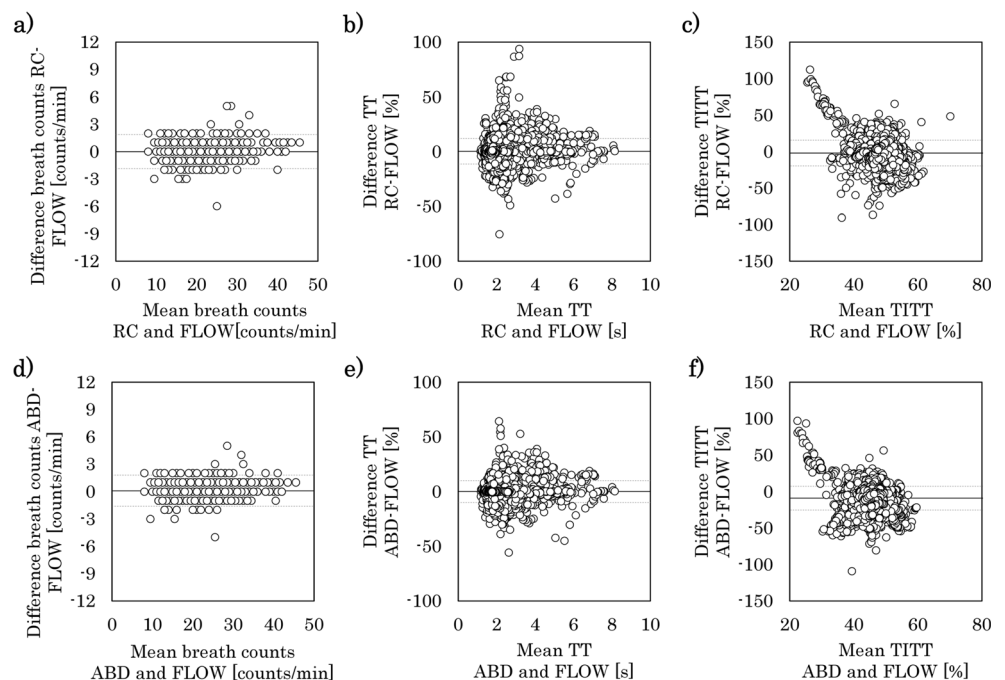
LOA limit of agreement, STS sit-to-stand/stand-to-sit, TT total duration of the respiration cycle, TITT inspiration time ratio

observed throughout the test protocol in healthy participants, whereas TITT detection was less accurate.

Differences in breath counts between STR and FLOW were small and within the predefined LOE. The result indicates that the new system successfully distinguished respiratory thoracic expansion from other motion artifacts when measuring respiration from thoracic movement during walking. In this experiment, three types of motion artifacts were implemented in the protocol: periodic leg and arm movements by walking, irregular and asymmetric movements during stick

walking, and slow postural changes during standing up and sitting down. Many were successfully discriminated with an algorithm that uses past respiratory cycles to search for local extrema. The accuracy was maintained regardless of cycles and volume of respiration. This may be because the respiratory cycle, generally less than 1 Hz, was much less than the walking cycle even at a rate of 1.5 km/h. In the 6-min walking test, a representative field walking test, most healthy adults achieved more than 200 m per 6 min, equivalent to 2.0 km/h [12]. Previous study using other wearable strain sensors

**Fig. 6** Bland-Altman plots of **a, d** breath counts; **b, e** total duration of respiration cycle (TT); and **c, f** moving average of inspiration time ratio (TITT), recorded by stretchable strain sensor (STR) vs corresponding values by analog signal of respiratory flow (FLOW) obtained by breath-by-breath in all participants. Upper panels (**a, b, c**) show data from the ribcage (RC) sensors, and lower panels (**d, e, f**) from the abdomen (ABD) sensors. Solid and dashed horizontal lines represent the mean difference (bias) and the limits of agreement (mean  $\pm$  1.96 SDs), respectively



**Table 5** The difference of inspiration time ratios between flow and stretchable strain sensors by movements conditions in relative difference

Sensing part	Condition	<i>n</i>	Breaths Counts	Bias %	SD %	Lower LOA (%n)		Upper LOA (%)	
						Mean	95% CI	Mean	95% CI
Ribcage	Standing	24	2536	-1.5	11.5	-24.5	[-31.4, -17.5]	21.5	[14.5, 28.5]
	Sitting	24	1761	-3.8	7.0	-17.8	[-22.0, -13.5]	10.1	[5.9, 14.3]
	Walking	24	3586	-3.6	9.1	-21.9	[-27.4, -16.4]	14.6	[9.1, 20.2]
	Stick walking	24	3967	2.2	7.0	-11.8	[-16.1, -7.6]	16.2	[11.9, 20.4]
	STS	24	1538	-3.0	9.5	-22.0	[-27.8, -16.2]	16.1	[10.3, 21.8]
Abdomen	Standing	24	2543	-10.0	11.3	-32.6	[-39.4, -25.7]	12.6	[5.7, 19.4]
	Sitting	24	1760	-10.1	6.8	-23.8	[-27.9, -19.6]	3.6	[-0.5, 7.8]
	Walking	24	3597	-9.6	8.1	-25.7	[-30.6, -20.8]	6.5	[1.7, 11.4]
	Stick walking	24	3971	-5.6	6.0	-17.6	[-21.3, -14.0]	6.5	[2.8, 10.1]
	STS	24	1560	-9.0	8.8	-26.5	[-31.8, -21.2]	8.6	[3.3, 13.9]
Overall	Ribcage	24	13388	-1.9	8.8	-19.6	[-24.9, -14.2]	15.7	[10.4, 21.0]
	Abdomen	24	13431	-8.8	8.2	-25.2	[-30.2, -20.3]	7.6	[2.6, 12.5]

LOA=limit of agreement. STS= sit-to-stand/stand-to-sit.

reported less than 0.1 s of bias between the strain sensor and flow sensor in measuring respiratory cycles [8]. The result in this work was comparable to the previous study. Considering these findings and settings, the STR system can be used accurately in most situations of the field walk test.

The time detection of TITT showed larger LOA values and weak correlation to FLOW. With FLOW, the phase is switched at the time when the output crosses zero, while in STR, the phase switching is set at the local maximum value and the local minimum value. Changes in thoracic circumference during

complicated dynamics between increased lung volume and alveolar pressure that were modulated by viscoelasticity of muscle and soft tissues in the thoraces. More than 90% of TT error were smaller than predefined LOE but the LOA of RC slightly exceeded the boundary. When comparing FLOW and STR by separating expiration and inspiration, the difference increased. This was agreed with RIP studies [14]. Motion artifacts sometimes modulated or masked the sharp peak of local extrema which directly influenced phase detection. For more accurate phase determination, it is necessary to implement a numerical

**Table 4** Total duration of respiratory time comparison between flow and stretchable strain sensors by conditions in relative difference

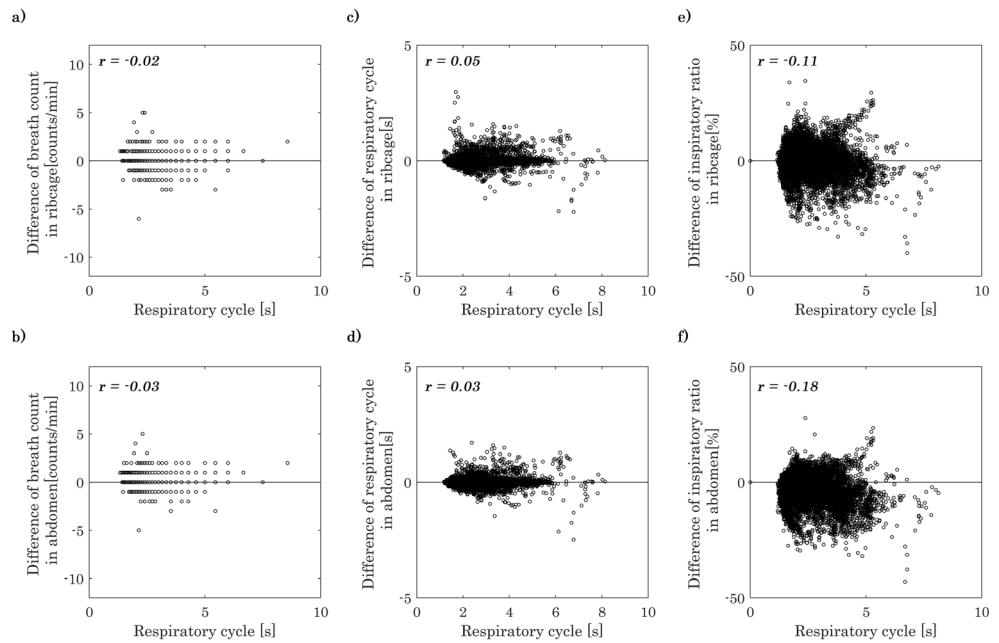
Sensing part	Condition	<i>n</i>	Breaths Counts	Bias %	SD %	Lower LOA (%)		Upper LOA (%)	
						Mean	95% CI	Mean	95% CI
Ribcage	Standing	24	2536	0.96	7.14	-13.3	[-13.7, -12.91]	15.24	[14.84, 15.64]
	Sitting	24	1761	0.72	4.82	-8.9	[-9.3, -8.59]	10.37	[10.04, 10.70]
	Walking	24	3594	-0.35	5.09	-10.5	[-10.8, -10.30]	9.83	[9.59, 10.07]
	Stick walking	24	3967	0.29	8.54	-16.8	[-17.2, -16.41]	17.38	[16.99, 17.76]
	STS	24	1538	1.54	6.73	-11.9	[-12.4, -11.43]	15.00	[14.51, 15.49]
Abdomen	Standing	24	2543	0.59	6.04	-11.5	[-11.8, -11.15]	12.67	[12.32, 13.01]
	Sitting	24	1760	0.54	4.56	-8.6	[-8.9, -8.28]	9.67	[9.36, 9.98]
	Walking	24	3606	-0.75	4.89	-10.5	[-10.8, -10.30]	9.03	[8.79, 9.26]
	Stick walking	24	3971	-0.09	4.36	-8.8	[-9.0, -8.61]	8.62	[8.43, 8.82]
	STS	24	1560	0.54	6.07	-11.6	[-12.0, -11.16]	12.68	[12.24, 13.11]
Overall	Ribcage	24	13396	0.63	6.47	-12.3	[-12.7, -11.93]	13.56	[13.19, 13.93]
	Abdomen	24	13440	0.16	5.18	-10.2	[-10.5, -9.90]	10.53	[10.23, 10.84]

LOA limit of agreement, STS sit-to-stand/stand-to-sit

breathing are not proportionally reflecting flow or volume change. The change of thoracic circumference is a result of

model to infer masked peak under the noisy situation.

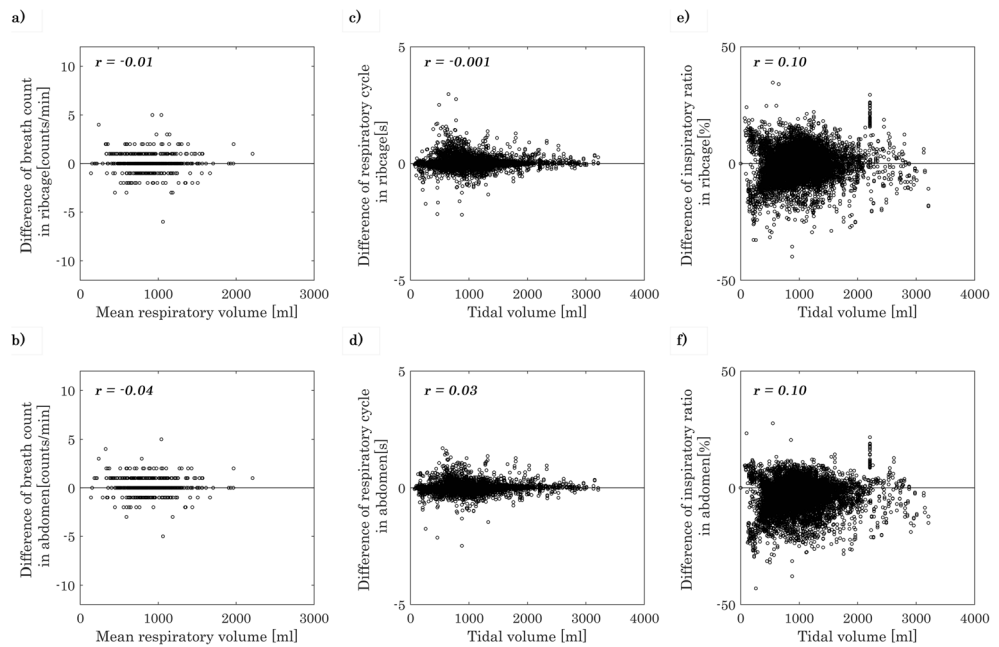
**Fig. 8** Scatter plots to show total duration of respiratory cycles and difference measurements of **a, d** breath counts; **b, e** total duration of respiratory cycle; and **c, f** moving average of inspiratory ratio, recorded by stretchable strain sensor vs corresponding values by respiratory flow sensor in all participants. Solid horizontal line represents the value of zero. Breath counts were calculated in every minute. Total duration of respiration cycle and inspiratory time ratio were measured by breath-by-breath. Values of italic “r” represent coefficients of Pearson’s product moment correlation analysis



Nowadays, there has been several studies reporting wearable respiratory monitor using strain sensors. Comparing to the studies, the system in this study The confidential intervals of biases of TT included zero, which was equivalent to the result reported by a study using resistive-type strain sensors [8]. Comparing to the previous studies the novel system has two advantages to measure respiration [8, 22–24]. First, the current system using a band with two sheets of the strain sensors is robust to motion artifacts as compared to directly skin-mounted small sensors or a single sensor waistband [8,

22]. This is because small sensors were highly sensitive to local deformation and placement of sensors largely affected signals. Our system placed two sensors symmetrically and covered a 200-mm length of torso movements to reduce motion artifacts. Second, our strain sensor system can measure large chest expansion. Chest expansion is ranged from 10 to over 100 mm in a healthy population [24]. The sensor in this study expands 100 mm each; thus, the chest-band system was designed to measure up to 200 mm of chest expansion. The stretchability is based on the characteristics of capacitive-type

**Fig. 7** Scatter plots to show tidal volume and differences of **a, d** breath counts; **b, e** total duration of respiratory cycle; and **c, f** moving average of inspiratory ratio, recorded by stretchable strain sensor vs corresponding values by respiratory flow sensor in all participants. Solid horizontal line represents the value of zero. Breath counts were calculated in every minute. Total duration of respiration cycle and inspiratory time ratio were measured by breath-by-breath. Values of italic “r” represent coefficients of Pearson’s product moment correlation analysis



**Table 6.** Effects of physical factors to accuracy of respiratory measurements

(n = 24)		Breath count		TT		Inspiration time ratio	
Correlation to RMSRD		rho	P	rho	P	rho	P
Ribcage	Chest expansion	0.22	0.304	0.03	0.90	- 0.07	0.74
	BMI	- 0.3	0.159	0.06	0.78	0.21	0.33
Abdomen	Chest expansion	0.27	0.20	- 0.04	0.87	0.03	0.87
	BMI	- 0.36	0.082	0	0.99	- 0.05	0.83
Sexual difference		MedD [95%CI]	P	MedD [95%CI]	P	MedD [95%CI]	P
Ribcage		- 0.49 [- 1.9, 0.80]	0.55	0.62 [- 1.7, 2.3]	0.51	- 0.42 [- 4.2, 2.9]	0.80
Abdomen		0.11 [- 1.17, 1.2]	0.93	0.79 [- 0.65, 1.9]	0.20	1.09 [- 1.5, 7.6]	0.27

\*MedD: median of difference. BMI = body mass index. RMSRD = root mean squared relative difference. CI = confidential interval. TT = total duration of the respiration cycle.

strain sensor showing relatively large stretchability as compared to resistive-type sensor based on the fabrication theory [1]. However, whole disagreement of the strain sensor reported in previous articles was unclear because the limits of agreement were not reported [8, 23, 24, 35]. This is the first study reporting the whole accuracy data of wearable strain sensors in breath counts, respiratory cycles and phase. Further studies should set an appropriate design to enhance clinical application of wearable strain sensors.

Comparing to other available monitoring methodologies, currently suggested system achieved substantial level of monitoring accuracy in breath counts and TT. Traditionally, respiratory inductive plethysmography has been used to measure time and respiration frequency from the thoracic circumference [6, 10, 15]. Clarenbach et al. [10] reported that TT for healthy subjects obtained from a respiratory inductive plethysmograph built into clothing is accurate with LOA less than 5% to flowmeter values. However, only 80 breaths in a stable breathing pattern were evaluated and transitions between exercise conditions were not included. In moderate exercise conditions, respiratory inductive plethysmograph had larger LOAs (- 2.6 and 3.2 bpm) [6] than those in our study. Johnstone et al. used an inertial measurement sensor based breath frequency measurement device [18], and reported a similar mean bias within 1 breaths/min from rest to treadmill walking at 6 km/h, with correlation coefficients of  $r = 0.81 \sim 0.84$ . Because the inertial measurement sensing system measures trunk movements at only one circumference, the accuracy and reliability of STR may be slightly better than that of the system. This study evaluated the accuracy of the STR system in a variety of posture and walking speeds and manners, and showed good agreement to FLOW sensor. Comparing other comparative devices, the breath counts and TT might be useful in rehabilitation programs or the field walking tests.

The experiment revealed that sexual and physical factors including chest expansion did not affect the accuracy of breath indices. This supports the both RC and ABD showed similar extent of the agreement to FLOW in

breath count. On the other hand, in TT ABD sensor demonstrated slightly smaller LOA for TT by about 2% as compared to RC. This may be because chest expansion at the abdomen was larger than ribcage leading better signal to noise ratio to search local extrema from the signal. However, the cause of noise during walking differs between sites: leg movement and postural change by sitting/standing largely affected the signal of ABD, but arm swing is the main noise source in the ribcage sensor generating more complicated. Thus, selection of a sensing site should carefully consider body movements during sensing.

In the present study, we successfully measured respiratory cycles and breath counts with stretchable strain sensors in healthy participants during moderate exercise. The advantages of this system are its non-invasiveness, its unconstrained measurements, and the simplicity of setup. Our goal is the clinical application of the system for real-time monitoring of respiratory rate in rehabilitation programs including field walking tests. The current calculation algorithm is designed for real-time monitoring. After a few seconds of storing signal data, the system starts respiratory phase detection and gives breath number, respiratory cycles, and phase information instantaneously. While a wireless transmission system is still needed to be developed, the findings of this study opens the possibility for real-time monitoring of respiratory rate in rehabilitation programs. However, our study has some limitations. The participants were healthy young adults; patients with chronic respiratory diseases generally have limited chest expansion compared to that of healthy individuals [20]. Although we report no significant association between chest expansion measurements and STR accuracy, the validity of this method should be investigated in patients with limited chest expansion, including post-operative patients and patients with chronic obstructive pulmonary disease. Moreover, our study focused only on time domain measurements of breathing rather than on special measurements such as tidal volume. The agreement with existing methods and the real-time detection ability of the STR system that we describe would benefit from

a recently advanced machine learning approach [11] to provide estimates of volume measurements. The third limitation of our study was the slow maximum walking speed relative to that of previous studies [10, 18]. Applicability of the current prototype of our system to activities such as sports or running should be investigated with another experimental protocol.

## 5 Conclusions

In conclusion, the new respiratory monitoring system achieved sufficient accuracy for clinical application with less than 3 counts/min difference of breath counts and less than 14% error of respiratory cycles in young healthy participants walking at variety of speeds and walking manners. By contrast, the respiratory phase detection is beyond the limit of clinical application. The wearable monitoring system equipping the stretchable sensor may be applicable as a real-time respiratory monitoring system in rehabilitation programs especially in the field-walking test.

**Author contributions** The literature search was done by AY, YO, YG and YW; data collection by AY, YB, YO, YF, YW, and KO; and all authors contributed to study design. Data analysis was done by AY, manuscript preparation by AY, and manuscript review was done by all the authors.

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## Compliance with ethical standards

**Conflict of interest** AY, HN, TT, AI received research grants and part of the experimental materials used for data collection from Bando Chemical Industries Ltd. YB is an employee of Bando Chemical Industries, Ltd.

**Ethical approval** All procedures were in accordance with the ethical standards of the institutional and/or national research committee and with the 1964 Helsinki declaration and its later amendments. The study was approved by the ethical standards of the Committee on Human Experimentation at the Graduate School of Health Sciences, Kobe University.

**Informed consent** Individual written informed consent was obtained from all participants included in the study.

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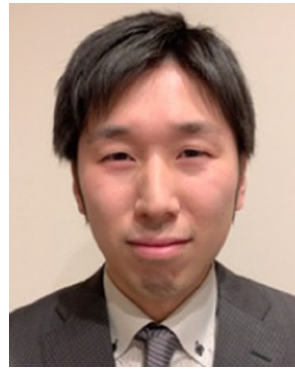
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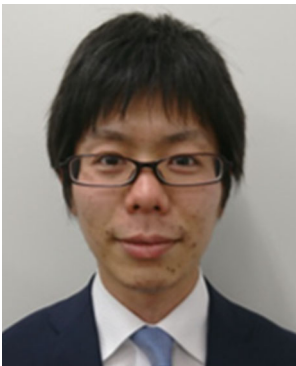
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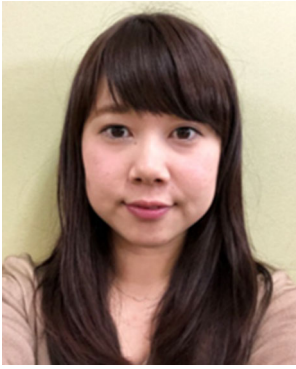
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