Noncontact Vital Sign Monitoring System with Dual Infrared Imaging for Discriminating Respiration Mode

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Abstract  Oral respiration causes constriction of the upper airway in the retropalatal and retroglossal regions, thereby increasing the risk of sleep disorder. One of the best methods to detect early signs of obstructive sleep apnea syndrome (OSAS) is daily monitoring of the respiration rate and mode of respiration during sleep. The vital signs are measured by a noncontact method in order to avoid burdening the subject and to allow differentiation between the various respiratory modes. In this study, we developed a system to measure the respiration rate and mode using far- and near-infrared cameras, and assessed the effectiveness of the proposed system and algorithm. A near-infrared camera detected the positions of the nostril and mouth, which are the pathways of expired and inspired air, respectively; while the far-infrared camera measured temperature changes in the nostril and mouth to derive the respiration rate and mode for detecting apnea. We enrolled 10 participants and measured their respiration rates using the aforementioned system under three states: nasal respiration, oral respiration, and apnea. The root-mean-square error for the respiration rate was 0.27 bpm, indicating that the system measured respiration without error in 92% of the trials. There was no error in discriminating between nasal and oral respiration. Additionally, this system detected apnea quite satisfactorily. The results of the experiment confirm that the system we developed effectively measures respiration in a noncontact manner.

Keywords: obstructive sleep apnea, dual infrared camera, respiratory method.

1. Introduction

Lifestyle parameters such as diet, exercise, and sleep have significant effects on our health. Sleep deprivation, in particular, may increase the burden on the heart, which leads to arrhythmias (such as atrial fibrillation) and other complications (such as hypertension and coronary artery disease) [1, 2]. Sleep apnea syndrome (SAS) is a typical sleep disorder. There are two main types of sleep apnea syndrome: obstructive sleep apnea syndrome (OSAS) caused by obstruction of the upper airway, and central sleep apnea syndrome (CSAS) caused by the failure of proper transmission of signals from the brain to the muscles that control breathing. OSAS contributes to the progression of heart failure [3]. In acute OSAS, the lack of oxygen causes ischemia and arrhythmia, and in the chronic stage, OSAS may lead to left ventricular hypertrophy and heart failure, with increase in mortality [4]. In the US, the prevalence of OSAS in adults is estimated to be approximately 20% [5]. On the other hand, an estimated 90% of people with moderate to severe OSAS are undiagnosed [6]. The presence of obvious symptoms does not necessarily lead to a correct diagnosis, which implies that there is very little awareness about OSAS, hence highlighting the need for early detection.

In CSAS, the respiratory rate is useful for the detection of sudden apnea due to failure of proper signal transmission. On the other hand, since the respiratory pathway is involved in worsening of symptoms in OSAS, it is necessary not only to estimate the respiratory rate but also to discriminate the respiratory mode. This is because OSAS is mainly caused by upper airway obstruction in the retropalatal and retroglossal regions [7], and oral respiration significantly constricts these regions [8]. Ob-
struction is also caused by obesity; conversely, disrupted metabolism due to sleep disorders exacerbates obesity [9]. Patients with OSAS spend more time breathing through their mouths than those who simply snore [10]. In addition, it is known that oral breathing occurs before/soon after apnea/hypopnea and that the appearance of mouth breathing triggers a vicious cycle that further increases apnea/hypopnea, which in turn leads to more frequent oral breathing [10]. In other words, the increase in mouth breathing and the severity of OSAS interact with each other. Therefore, estimation of respiratory rate to detect apnea in CSAS and OSAS, together with discrimination of respiratory mode will be useful to understand the process of worsening and improvement of OSAS. This approach is also expected to contribute to preventive medicine.

It is desirable to measure respiration on a daily basis in order to monitor the processes of worsening and improvement due to bad lifestyle habits and behavior modification, respectively. The airflow sensor currently used for simple diagnosis of SAS requires the patient to wear a measuring device, which is burdensome for the patient and unsuitable for daily measurement. Therefore, a non-contact instrument is needed to measure vital signs for detecting sleep apnea without burdening the patient unnecessarily.

In previous studies, devices for non-contact measurements have been developed to measure vital signs in various situations in daily life [11]. In particular, while sleeping, a few physiological indices may be measured by examining body movements and estimating the respiratory rate. The changes in pressure applied on the bed by the human body while sleeping is determined [12–14]. In noncontact methods, the body movement and respiration rate are estimated by measuring the expansion and contraction of the chest during respiration using a depth sensor, or by measuring the temperature change of the nasal cavity while sleeping using a far-infrared camera [15–17]. However, in each case, only the respiration rate is estimated, and thus far, little attention has been paid to the technique that distinguishes between respiration modes.

To determine the respiration mode, it is necessary to measure data focusing on the differences between the various modes. The most suitable one is to measure skin temperature because the region of temperature change is different in nasal and oral respiration. The temperature of expired air is almost equal to that of the core (34–36 °C) [18, 19], while that of inspired air is the same as room temperature (23–28°C). Compared with facial skin temperature (32–34°C) [20], expiratory air is warmer, and inspiratory air is cooler (Fig. 1). Therefore, the nostril and mouth temperatures change cyclically along with expiration and inspiration. These cyclic temperature changes provide an index of the respiratory mode, which may indicate the risk of OSAS.

This study focused on the variation in the pathway of expired and inspired air in different modes of respiration, and measured changes in skin temperature caused by this variation. The purpose of our study was to develop a system to measure the respiration rate and respiration mode using far- and near-infrared cameras, and to assess the effectiveness of the proposed system and algorithm. Experiments were performed to measure skin temperature under three conditions: nasal respiration, oral respiration, and apnea.

2. Methods

2.1 Measurement

To measure temperature, it is necessary to perform thermal imagery of the nose and mouth. However, it is difficult to automatically detect them while using a far-infrared camera. The device we developed consists of a near-infrared camera and a far-infrared camera so that it can also be used while sleeping in the dark. In this study, we used an ARTCAM-036MI-WOM near-infrared camera (60 fps at 640 × 480; Artray, Tokyo, Japan) to detect and apply the coordinate data of each area to the image obtained using an OWLIFT Type-A far-infrared camera (8.6 fps at 80 × 60; Infinitegra, Kanagawa, Japan) (Fig. 2). Figure 3 shows the outline of the measurement device used. In addition, we used an OSFKSSCS near-infrared light-emitting diode (LED) (λpeak = 850 nm; OptoSupply, Hong Kong, China) as the light source in the
dark.

### 2.1 Signal Processing

The near- and far-infrared images have different numbers of pixels and angles of view. Therefore, it was necessary to associate the coordinates of the images. **Figure 4** shows how the near- and far-infrared images are associated. From the equation of the ratio of the distance and that of the number of pixels, Eqs. (1) – (4) are derived as follows:

\[
L \tan \theta_h : L \tan \frac{\theta_{nir,h}}{2} = \frac{w_{nir}}{2} : \frac{x_{nir}}{2}, \quad (1)
\]

\[
L \tan \theta_h : L \tan \frac{\theta_{fir,h}}{2} = \frac{w_{fir}}{2} : \frac{x_{fir}}{2}, \quad (2)
\]

where \(x_{fir}\) and \(x_{nir}\) are the x-coordinates of the two cameras, and \(\theta_{nir,h}\), \(\theta_{fir,h}\), and \(\theta_h\) are the horizontal angles of view of the two cameras and \(x_{nir}\), respectively. \(w_{fir}\) and \(w_{nir}\) represent the horizontal resolution of the two cameras, which are 80 pixels and 640 pixels, respectively. The distance from the system to the participant is designated \(L\).

\[
L \tan \theta_v : L \tan \frac{\theta_{nir,v}}{2} = \frac{h_{nir}}{2} : \frac{y_{nir}}{2}, \quad (3)
\]

\[
L \tan \theta_v : L \left( \tan \frac{\theta_{fir,v}}{2} - \frac{d}{L} \right) = \left( \frac{h_{fir}}{2} - \frac{dh_{fir}}{2L \tan \frac{\theta_{fir,v}}{2}} \right) : \frac{y_{fir}}{2}, \quad (4)
\]

where \(y_{fir}\) and \(y_{nir}\) are the y-coordinates of the two camera, and \(\theta_{nir,v}, \theta_{fir,v},\) and \(\theta_v\) are the vertical angles of view of the two cameras and \(h_{nir}\), \(h_{fir}\) and \(h_{nir}\) represent the horizontal resolution of each camera, which are 60 pixels and 480 pixels, respectively. The difference between the optical axes of the two cameras is designated \(d\).

Using these equations, the coordinates of the far-infrared image can be obtained from those of the near-infrared image using Eqs. (5) and (6) as follows:

\[
x_{fir} = \frac{w_{fir}}{w_{nir}} \tan \frac{\theta_{fir,h}}{2} \left( \frac{x_{nir}}{2} - \frac{w_{nir}}{2} \right) + \frac{w_{fir}}{2}, \quad (5)
\]

\[
y_{fir} = \frac{h_{fir}}{h_{nir}} \tan \frac{\theta_{fir,v}}{2} \left( \frac{y_{nir}}{2} - \frac{h_{nir}}{2} \right) - \frac{dh_{fir}}{2L \tan \frac{\theta_{fir,v}}{2}}, \quad (6)
\]

We detected 68 face landmark points from the near-infrared image using Dlib, an open-source software library, and applied the coordinate data to the far-infrared image in order to create a region of interest (ROI). **Figure 5** shows the annotation numbers of the face landmarks and two ROIs calculated using Eqs. (7) and (8), as follows:

\[
ROI_{nasal} = \begin{pmatrix}
X \\
Y \\
H \\
W
\end{pmatrix} = \begin{pmatrix}
x_{32} \\
y_{31} \\
x_{36} - x_{32} \\
y_{34} - y_{31}
\end{pmatrix}, \quad (7)
\]
where $X$ is the horizontal coordinate of the starting point in the ROI, $Y$ is the vertical coordinate, $H$ is the height, and $W$ is the width. $x_i$ and $y_i$ represent the coordinates of the $i$th facial point in Fig. 5.

Skin temperature must be expressed relative to ambient temperature in order to avoid the effect of room temperature and the heat of the thermal camera itself. We recorded the intensity using Eq. (9) as follows:

$$s_r(k) = \frac{1}{HW} \sum_{i=X}^{X+H} \sum_{j=Y}^{Y+W} s(i, j, k) - \bar{S}(k),$$

where $s(i, j, k)$ is the intensity on the coordinate $(i, j)$ at the $k$th far-infrared video frame; $s_r(k)$ is the difference between skin temperature, averaged spatially over all pixels in the ROI, and $\bar{S}(k)$ is ambient temperature averaged spatially over all pixels in the far-infrared image. We smoothed the signal using a low-pass filter with a cutoff frequency of 0.145 Hz.

### 2.2 Algorithm

In a 80-second respiration measurement exercise performed prior to the experiment, measurements were taken under nasal respiration for the first 30 s, and under oral respiration for the next 30 s. To simulate the apnea state, the participant held his breath for the last 20 s. To simplify the detection of a change in temperature due to respiration, we differentiated and scaled the signals to transform the data such that the signals are within a range of $[-1, 1]$, using Eq. (10) as follows:

$$s_{d, sc}(k) = \frac{2(s_d(k) - s_{d, min})}{s_{d, max} - s_{d, min}} - 1,$$

where $s_d(k)$ is the differentiated intensity at the $k$th far-infrared video frame; $s_{d, max}$ and $s_{d, min}$ are the maximum and minimum intensity in the signal. The scaled signal $s_{d, sc}(k)$ increases and decreases during expiration and inspiration, respectively. Figure 6 shows how respiration can be detected. We set three thresholds to detect the respiration.

The signal first increases passing the first threshold (high signal: exhalation), then plateaus and decreases below the second threshold (low signal: inhalation); after reaching the trough, the signal increases again to exceed the third threshold (zero). Detection of the three thresholds is counted as one respiration.

The high and low thresholds were obtained according to the average signal intensity. Since there is almost no sudden drop in temperature except for inhalation, the average signal intensity is used for the high threshold. On the other hand, the low threshold needs to be adjusted to separate inhalation from other fluctuations. The parameter $\alpha$ in Figure 6 was determined empirically for proper analysis, which was set at 0.33 and 0.2 for the nostril and mouth, respectively. Respiration was recorded only when a discernible change in temperature was observed, because the differential signals sometimes emphasized noise. Figure 7 shows how to distinguish between the temperature change due to respiration from minor temperature fluctuations.

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perature change in 5 s, which is the time an adult takes one breath [21]. Figure 8 shows how to classify breathing modes. For the temperature changes measured in the nasal and oral ROI, the system distinguished between temperature changes due to breathing and slight fluctuations. If the temperature change in the nose was due to respiration, while the temperature change in the mouth was a fluctuation, the respiration was evaluated as nasal breathing. On the other hand, when temperature change due to respiration was observed in the mouth, the respiration was evaluated as mouth breathing, regardless of whether the nose showed temperature change due to respiration or fluctuation. One of the reasons why the nose also shows changes in temperature during mouth breathing is that the air flow through the nostrils is caused by the negative alveolar pressure with respect to the atmospheric pressure. Therefore, if the nasal and oral temperatures had similar cyclic changes, respiration is recorded as oral respiration. This situation occurs when the subject has nasal congestion. Detection of an apnea state was based on the definition of an apnea event. Thus, if no breaths were detected for 15 s from the end of the last inspiration, the system determined that the person was in an apnea state.

2.3 Experimental Conditions
We conducted an experiment to confirm whether the system was able to calculate the respiration rate as well as distinguish between respiration modes (nasal respiration, oral respiration, and apnea). We verified whether the respiratory rate and mode detected by the system matched those of the participant’s self-report as reference. Figure 9 shows the experimental setup. A participant leaned back on a chair at a distance of approximately 0.7 m from the system, with the camera set at an angle of approximately 30 degree to the participant’s face. The nose was positioned in the center of the image, and 68 facial landmarks were included in the image. Although the noses had different shapes, each nostril was recorded via a far-infrared video. The brightness of the room can be adjusted from dark (nearly 0 lx) simulating nighttime sleep, to bright (about 1000 lx) simulating daytime naps. This experiment was conducted under the bright condition.

The room temperature was 24.4°C (SD 0.43), and did not change rapidly during the experiment. The emissivity of the facial skin was set as $\varepsilon = 0.98$ based on previous studies [22, 23].

2.4 Participants and Procedure
We enrolled 10 participants (1 female and 9 males) with an average age of 23.3 years (SD 1.6). They were all healthy, and none of them had a sleep disorder.

The study was approved by the Faculty of Engineering, Information and Systems, Research Ethics Committee of University of Tsukuba. The experimental purpose and procedure were explained to the participants, and all provided informed consent. Figure 10 shows the experimental procedure.

1. Each participant was asked to close his or her eyes and rest on the chair for 5 min before measurement, in order to stabilize his or her condition.
2. For the first 30 s, measurements were taken under nasal respiration. The participant self-reported by pressing button 1 every time when inhaling through the nose.
3. For the next 30 s, oral respiration was performed. The participant self-reported by pressing button 2 every time when inhaling through the mouth.
4. To simulate the apnea state, the participant held his/her breath and pushed button 1 for the final 20 s.

During measurement, each participant breathed at a pace that he or she felt comfortable in a resting state. The above procedures were repeated five times.

First, from the measurements obtained from the 10 participants, the process of the proposed algorithm was
verified. Subsequently, the accuracy of the measurements was evaluated. Evaluation of the proposed algorithm focused on two points: (1) the possibility of calculating the respiration rate for 80 s during each respiration trial, and (2) the possibility of detecting the apnea state in the last 20 s of the trial.

These evaluations were conducted by comparing the results obtained using our system with the participants’ reports as reference. The accuracy of the respiratory rate calculation was evaluated by the root-mean-square error (RMSE) between the respiratory rate calculated by our system and the reference respiratory rate. The ability to determine apnea correctly was evaluated based on the capability of our system to detect no respiration, and thus determine that the final 20-second period of the experiment was the apnea state.

3. Results

3.1 Process of Algorithm

Figure 11 shows changes in nasal and oral temperatures obtained from the 80 s respiration measurement exercise. We observed a cyclic change in oral temperature during 30 s of oral respiration, and in nasal temperature during 30 s of nasal respiration. In the apnea state, no cyclic change in temperature was observed during respiration, showing that no air flow passed through the nostril and mouth. In contrast, oral temperature gradually increased up to a certain point after shifting to the apnea state. Since the participant breathed through the mouth just before the apnea state, the equilibrium state of the oral temperature became lower compared to that during the apnea state, because of the inflow of external air by inspiration. Therefore, oral temperature increased towards equilibrium at the beginning of apnea. During the apnea state, no air flow passed through the nostril and mouth; as a result, there was no cyclic change in temperature.

Figure 12 shows the process of distinction between the respiration mode. To simplify the detection of a change in temperature due to respiration, we differentiated the signals. Figure 12(a) and (b) show normalized signals of nasal and oral temperatures, respectively, after all the filters were applied. As shown in Fig. 12(c), respiration was recorded only during the period when the change in temperature was discernible. Figure 12(d) shows the method that detects oral respiration with a decrease in nasal temperature due to inspiration. If the inspiratory period is similar in nasal and oral respiration, it is recorded as oral respiration.

3.2 Respiratory Rate and Mode

When this algorithm was applied to the waveforms of temperature changes obtained in the experiment, the error between the respiration rate calculated by the system and the reference rate was 0 breaths per minute (bpm) in 46 of 50 trials, and 1 bpm in the remaining 4 trials. The root-mean-square error (RMSE) was 0.27 bpm. In terms of discriminating the respiratory mode in all trials, nasal breathing was not misidentified as mouth breathing, and mouth breathing was not misidentified as nasal breathing at all. In addition, the apnea state was detected by the system correctly (counting no breaths in the final 20 s) in 47 of 50 trials.

Figure 13 shows typical examples of temperature change in the mouth and nostril obtained in the experiment. The waveform in Fig. 13(a) was similar to the waveform measured in the preliminary experiment (Fig. 11), and calculation of the respiration rate, determination of the respiratory method, and detection of the apnea state were performed without error. Figure 13(b) shows a typical waveform in which a slight change in nasal temperature is observed during oral respiration. Even for such waveform, we found no error in those tasks. In contrast, Fig. 13(c) shows one of the four trials that resulted in an error in those tasks. As far as apnea detection is concerned, the system detected no respiration and determined the final 20 s of the experiment as the apnea state in 47 successful trials. However, in 3 trials of apnea, the minor temperature fluctuations independent of respiration were erroneously considered to be a result of respiration, leading to inaccurate respiration rates. As shown in Fig. 13(c), the temperature increased significantly at the last expiration before apnea, and the first temperature drop that occurred in the apnea state was detected as respiration.

4. Discussion

In this study, we applied the proposed algorithm to the thermal data of the nose and mouth based on the ROIs
from the near-infrared camera. This algorithm used two thresholds and smoothed SD to distinguish between the temperature change due to respiration from the minor temperature fluctuations detected by the far-infrared camera. As shown in Figs. 12(a) and (b), the two thresholds prevented the error in counting breaths due to fluctuations. Even when two troughs appeared during the temperature drop due to weakening of inspiration during breathing, such as the fourth nasal breath in Fig. 12(a), the algorithm detected only one breath. Moreover, as shown in Fig. 12(c), period with temperature fluctuations, such as the insignificant changes in oral temperature in the first 30 s, were detected by the smoothed SD as period without respiration. In fact, as shown in Fig. 12(b), the fluctuation in oral temperature in the latter half of the nasal respiration period crossed the two thresholds, with a possibility that it might be judged as a change due to respiration. However, based on the threshold shown in Fig. 12(c), it was determined that this change was not due to respiration. This process would make our system robust enough to withstand disturbances. Since the nose and mouth differ in the rate of temperature decrease due to breathing, it was necessary to determine whether the temperature decrease was caused by the same breath or not. In the proposed algorithm, if oral respiration was detected 2.5 s before and after the detection of nasal respiration, it was counted as a single oral respiration. In the experiment, this algorithm enabled us to prevent misdetection, as shown in Fig. 12(d), but the possibility of misdetection may increase if the breathing frequency is high.

The RMSE of 0.27 bpm, indicating measurement accuracy, is better than that of a commercial biological information monitoring device (< 2.0 bpm). Therefore, the respiration rate calculated by our system is sufficiently reliable. In addition, the system could distinguish between oral and nasal respiration without any error in all the trials, indicating that it may be possible to verify the percentage of oral respiration in the total respiration during sleep. The apnea detection time is the sum of 10 s (based on the definition of sleep apnea) and 5.0 s which is the time that an adult takes one breath [21]. Determin-
ing apnea is almost the same as counting breaths. Hence, apnea detection can be conducted as long as minor temperature fluctuations during apnea are not mistaken for respiration. Thus, the experimental results verified the efficacy of the proposed system and algorithm to measure the respiration rate and mode.

It is also possible to determine whether the nostrils are obstructed during oral respiration. Figure 13(a) and (b) show the differences between two typical results of the temperature changes in the nose during oral respiration. If no upper airway obstruction occurs, changes in the nasal temperature are observed during oral respiration. In contrast, nasal congestion will not be observed if the airway is blocked. Since the main cause of oral respiration leading to OSAS is upper airway obstruction [7] (as shown in Fig. 13(b)), the result demonstrated may be closer to that observed in real patients. However, some mouth-breathers habitually breathe through their mouths even without clear signs of airway obstruction [24]. Such mouth-breathing people may show results as shown in Fig. 13(a). The temperature of the nose changes even during the 30 s of oral respiration because of the air flow through the nostrils caused by the negative alveolar pressure relative to the atmospheric pressure, as mentioned before. Such change in temperature should be considered because upper airway obstruction does not always occur during sleep apnea. Determining whether the cause of oral respiration is upper airway obstruction using the proposed system may help SAS treatment.

In the experiment, we could not simulate a SAS patient turning over in bed during sleep. The measurement was performed under the condition that the system and the subject in supine position were facing each other. As the patient turns over during sleep, the relative position between the system and the face changes, and the nose and mouth may not be recorded on the video. On the contrary, even if a patient turns over and lies on his/her side, it may be possible to measure breathing if the nasal cavity and mouth are recorded in the far-infrared camera image. Additionally, it is expected that narrowing of the upper airway due to mouth breathing would occur more easily when the subject is in supine position. Therefore, it is important to observe respiration when the subject is in supine position.

Another concern is that we did not simulate cyclic motions of the chest or stomach caused by breathing efforts during apnea. The factor that affects measurement is a slight shift of the ROI in infrared image due to that motion. On the other hand, motion artifacts due to effort breathing vary from person to person, and is difficult to quantify. However, it is unlikely that this motion artifact would displace the mouth or nose from the ROI. In order to evaluate this effect, it is necessary to conduct an ex-

![Fig. 13](image-url) Typical examples of temperature changes in the mouth and nostril. (a) A typical result with no error. (b) Nasal obstruction during measurement. (c) A misdetection was caused by a little fluctuation during apnea.
periment on patients with SAS.

Previous studies have reported the feasibility of measuring the respiration rate by noncontact methods using a far-infrared camera [14–16]. Its accuracy depends largely on the setting of an appropriate ROI and an algorithm that extracts only respiration-induced temperature change. Depth cameras are often used to measure the expansion and contraction of the chest during respiration [14], but are more susceptible to body movements during sleep, and are also affected by duvets. Hence, far-infrared cameras are preferred for measuring respiration [14–16]. However, these studies ignored the methods of measuring respiration, which is closely associated with OSAS. The system we developed can calculate the respiratory rate and differentiate between the breathing modes accurately. These were achieved with the help of the respiratory detection algorithm, and the automatic detection of the ROIs (nose and mouth) from the near-infrared image.

In this study, we determined whether the breathing action of the participants was dependent on the changes in temperature of the nose and mouth. The result that the system can not only estimate the respiratory rate but also discriminate the mode of respiration suggests that it can be useful for both CSAS and OSAS screening. In addition, the system can be used to obtain the ratio of oral respiration to total respiration and the occurrence of apnea events using the far-infrared camera, and may also provide information on the patient’s posture while sleeping by observing with the near-infrared camera for daily monitoring.

5. Conclusion

We developed a monitoring system that can measure the change in temperature during each respiration cycle using near- and far-infrared cameras. The results obtained from the experiment with 10 participants verified that the system was able to capture the difference in temperature changes in the nose and mouth, depending on the respiratory state. We confirmed that the system had the potential to calculate the respiration rate as well as distinguish between oral and nasal respiration satisfactorily in a non-contact manner.

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Conflict of Interest

The authors declare no conflicts of interest with any companies or commercial organizations per the definition of Japanese Society for Medical and Biological Engineering.

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