ers that during the ejection period the ventricular elastance can be approximated to a linear curve and that the V_0 should be constant during this interval. Considering these two assumptions, V_0 is estimated by finding a linear approximation for the elastance during the ejection period that minimizes the variations of the V_0 calculated for each point in this period. The instantaneous elastance is calculated using the estimated V_0 and its maximum value is defined as the E_{max} (5). All measurements were obtained with open-chest condition. LVP was preprocessed with a zero-phase low pass filter ($F_0 = 10$ Hz), and LVV with a de-noising symlet wavelet (sym8). The measurements regarding the low cardiac function were started as soon as the aortic pressure stabilized after the propranolol injection. The statistical analysis of the results was based on the analysis of variance (ANOVA). A P value of less than 0.01 was considered statistically significant.

RESULTS

In most cases, even though the mean estimated values using the POM and the value estimated with the conventional method were different, there was no significant difference among the E_{max} values estimated with the POM for the same cardiac function with different rotational speeds. However, the values estimated for V_0 were not constant for each animal. Figure 1 shows the estimated values for E_{max} and V_0 for animal 1.

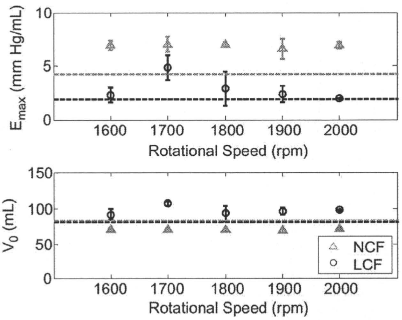

FIG. 1. (a) E_{max} and (b) V_0 comparing the conventional estimation method (dashed line) and POM (markers, mean ± standard deviation) for different pump rotational speeds using the data from animal 1. The black lines and markers represent the low cardiac function (LCF) and the gray lines and markers, the normal cardiac function (NCF).

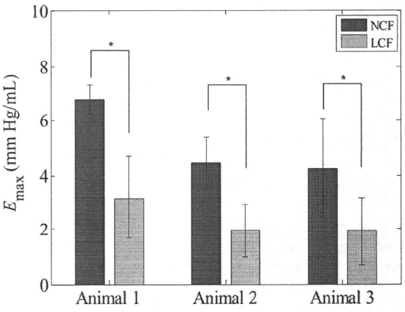

FIG. 2. E_{max} estimated for each cardiac function, independent of the pump rotational speed, represented by the mean and the standard deviation. *$P < 0.01$ between NCF and LCF.

After verifying that the E_{max} is independent of the pump rotational speed, the estimated values for the NCF and the LCF were compared. The results are represented in Fig. 2. It was found that there was a significant difference between the values for NCF and for LCF in the three animals.

DISCUSSION

E_{max} is considered the most reliable index for assessing myocardial contractility in the intact circulation and is almost insensitive to changes in preload and afterload (6). Further, it is an important cardiac function index for the detection of the recovery of assisted hearts with no need for pump stop, because it can detect cardiac function independently of the assistance level (4). In contrast, the EF, which is commonly used during the off-pump test, cannot be applied without stopping the pump due to its sensitivity to the afterload, which depends on the pump model and rotational speed.

Vanderberghe et al. analyzed the instantaneous elastance during the assistance with pulsatile and continuous flow pumps and concluded that the time-varying elastance model was not accurate during mechanical assistance (7). However, in that study the E_{max} was estimated using the V_0 obtained with the multiple beats method for the baseline condition, which is not consistent with recent findings that the V_0 might also change (3). Considering that changes in the pump rotational speed change the ventricular afterload out of the normal range of the circulatory system changes, V_0 is more likely to change depending on the pump rotational speed than the usual observation in unassisted hearts. Therefore, V_0 should also be continuously assessed, as in most single beat estimation methods.

Although some studies have shown that single beat estimation methods have insufficient precision (8), the results presented in this article indicate that there was no significant difference among the estimated values at each pump rotational speed, as shown in Fig. 1. This indicates that the E_{max} estimated with the POM is independent of the assistance rate, which is critical for the assessment of the recovery without stopping the pump and with no need for corrections on the index depending on the rotational speed.

The results also indicate that the value of V_0 changed within the data from the same animal. The increase in the estimated V_0 is a consequence of a combination of the increase of the pump assistance rate, the decrease of the aortic pressure, and the decrease of the ventricular contractility.

For the data corresponding to the NCF, in animal 1 there was an artifact on the LVP measurement caused by the shock of the conductance catheter on the left ventricular wall. This artifact led to the overestimation of the E_{max} for those data in comparison to the value estimated using the conventional method, as shown in Fig. 1.

Moreover, due to the invasive nature of placing the RBP and the open chest conditions, the cardiac function may have changed during the measurements. In the case of the induced low cardiac function, the cardiac function depends on the quantity of medicine and the duration of the measurements. In particular, in animal 1, after stabilizing for the measurement of the LCF at 1700 rpm, it was observed that the end systolic LVP increased around 7% with no other change in the pump rotational speed or any other external change. Consequently, another dose of propranolol was administrated before the measurement with the pump operating at 1800 rpm. The off-pump data was taken after this additional injection.

Finally, the E_{max} estimated for the control data and the E_{max} estimated for the condition mimicking heart failure were significantly different. This result was observed in the data from each of the three animals, as shown in Fig. 2. This indicates that the E_{max} estimated with the POM can detect changes in the ventricular contractility, and could potentially be used to assess the recovery. It is still necessary to evaluate the variations of the E_{max} during gradual changes in the cardiac function, mimicking the scenario of the bridge-to-recovery. There is also a need to find a correspondence between the cardiac function after the pump withdrawal and the E_{max} value during assistance with the RBP, defining a weaning criterion.

Some recent studies affirmed that due to the variations in V_0, both E_{max} and V_0 should be considered to assess the cardiac function, for example with analysis of covariance (3), which should be accounted in future studies.

Although E_{max} is an important cardiac function index, it has limited clinical applicability because the conventional multiple P-V loops method for assessing E_{max} requires sudden changes in the pre- or afterload. In this study, one of the single beat methods, the POM, was evaluated using animal experiment data with assistance of a centrifugal RBP; however, other single beat estimation methods have been proposed, based on a different feature of the elastance (5,9–11). It is therefore necessary to evaluate them in order to find which one fits the hemodynamics of the assisted heart better.

The main limitation of this study is that data from only three animal experiments were considered, which is not sufficient for a definitive conclusion. A detailed study is important because there is no guarantee that ventricular function during cardiac assistance with a RBP would be representative of what ventricular function would be without the device. A failing heart evaluated as recovered with the device still on, could actually not be able to provide sustained recovery over time.

In addition, the measurements of LVV and LVP are highly invasive when attained by placement of conductance catheter inside the left ventricle and are also sensitive to the correct catheter placement. In order to reduce invasive procedures, the measurement of LVV from the conductance catheter might be substituted by the total flow ejected from the ventricle. Such a method was proposed for the assessment of the cardiac function using the POM, in data obtained from unassisted hearts where there is only one flow from the left ventricle, and must be adapted to the case of assistance with the RBP when—besides the flow through the aortic valve—there is also blood flow through the pump.

CONCLUSION

In this article, we have discussed the adequacy of an index of cardiac function E_{max} that was estimated with the POM, a single beat estimation method, on the basis of in vivo experimental data obtained during RBP assistance. Although the estimation of E_{max} value was not precise enough to coincide with the true absolute values obtained by the traditional method in all conditions, the results indicate that the E_{max} is independent of the pump rotational speed and sensitive to changes in the cardiac function, which is fundamental to the detection of myocardium recovery.

Acknowledgment: The authors are most grateful to Kimio Kikuchi for the support during the animal experiments.

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Influence of Rotary Blood Pumps over Preload Recruitable Stroke Work

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***Abstract*— When recovery of the cardiac function is detected in assisted hearts, the ventricular assist device can be removed. Due to the invasiveness of the surgical procedure, an accurate assessment of cardiac function is fundamental for the treatment success. The main challenge for the detection of cardiac function during assistance is to know whether the cardiac function index represents the cardiac function after pump removal independently of the pump assist rate. Therefore in this paper we present an evaluation of the influence of the pump over the slope of the preload recruitable stroke work, a cardiac function index. Analyzing data from four acute animal experiments, we found that the pump affects the stroke work, which could be corrected by the end diastolic volume. However, the data set examined was limited and further investigation is necessary.**

I. INTRODUCTION

Cardiovascular diseases are the leading cause of death in the world [1]. In the past decades, new devices for the diagnostic and treatment of such diseases were developed. Small implantable rotary pumps are now used as ventricular assist devices supporting the blood circulation of heart failure patients waiting for a heart transplant (bridge-to-transplant) and also to temporarily unload the ventricle during treatments such as cell therapy (bridge-to-recovery). In a bridge-to-recovery, when the myocardium has recovered, the pump is withdrawn [2], [3], [4]. The correct detection of the ventricular recovery is fundamental for the success of the treatment due to the invasiveness of the procedure for pump removal and for an eventual reimplantation [5].

Approaches such as assistance rate reduction protocols have been empirically proposed [6], [7]. However, during the support of ventricular assist devices, there are considerable changes in the cardiovascular system; for example, if the pump rotational speed is high the coronary perfusion might not be sufficient even when there is enough blood being pumped in the systemic circulation. Therefore there is no guarantee that the assessed cardiac function corresponds to the cardiac function after the pump removal.

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Besides, the influence of the assistance over the index that will be used is also fundamental. There are many cardiac function indices used for the diagnostic of heart failure; however, the influence of the assistance with rotary blood pumps (RBPs) over those indices has not been clarified yet [8], [9], [10], [11]. Before using a cardiac function index for the assessment of the myocardial recovery, we should know the relationship between pump rotational speed and the index as well as verify if despite the assistance, the index is sensitive to changes in the ventricular contractility.

An index commonly used for the assessment of the cardiac function, also in unassisted hearts, is the preload recruitable stroke work (PRSW), which is independent of ventricular afterload [12].

Aiming at an optimal protocol for the assessment of the myocardial recovery during the assistance, the objective of this study is to evaluate the effects of the RBP over PRSW, in order to analyze whether it is a good candidate for the assessment of the recovery.

II. METHODS AND MATERIALS

In this study we used data from acute animal experiments (4 female adult healthy goats) with the centrifugal pump Evaheart LVAS (Sun Medical Res. Tec. Corp, Japan). The pump was connected in a bypass configuration with the inflow cannula inserted into the left ventricular apex and the outlet anastomosed to the descending aorta as represented in Fig. 1.

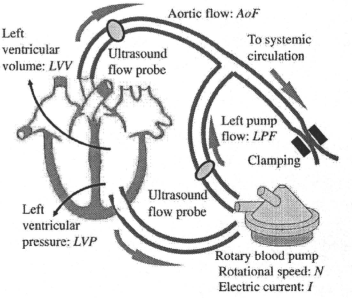
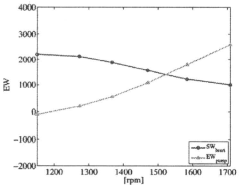
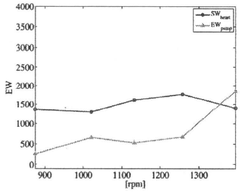


Fig. 1. Experimental Setup

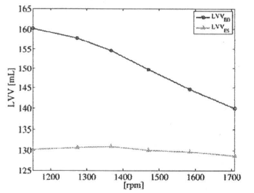


(a) Normal Cardiac Function

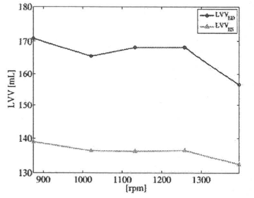


(b) Low Cardiac Function

Fig. 2. Changes in external work (EW) against pump rotational speed - Animal # 1



(a) Normal Cardiac Function



(b) Low Cardiac Function

Fig. 3. Change in left ventricular volume (LVV) against pump rotational speed - Animal # 1

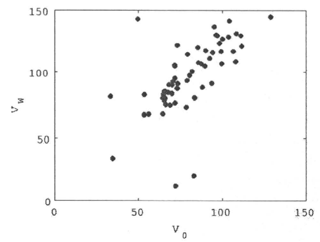


Fig. 4. Comparison of the x-axis intercept point of the PRSW (V_w) and of the ESPVR (V_0) estimated for data with aortic clamp

All measurements were obtained at open-chest condition. Data were recorded during 60s with at least four different assistance levels. Propranolol was administrated to induce a decrease in the ventricular contractility; data recorded before this injection corresponds to a normal cardiac function (NCF) and the data recorded after, to a low cardiac function (LCF). Left ventricular pressure (LVP) and volume (LVV) were measured by a conductance catheter (Leycom, Netherlands) inserted inside the left ventricle from the left atrium. In some parts of the data, we observed an artifact on the LVP caused by the shock of the catheter tip on the ventricular wall at the beginning of the systolic period. LVP was pre-processed with a zero-phase low pass filter (F_0=15Hz), and LVV with a de-noising symlet wavelet (sym8) [5].

Left pump flow (LPF) and aortic flow (AoF) were monitored with ultrasound flow probes (Transonic Inc., US) as reference for the pump rotational speed (N) control.

Stroke work (SW) was calculated as area inside the PV-loop of each cardiac cycle. The slope (M_w) of the PRSW, defined in (1), was used as an index of the cardiac function.

$$M_w = \frac{SW}{V_{ED} - V_w} \tag{1}$$

where V_{ED} is the end diastolic LVV and V_w is the x-axis intercept of the PRSW. For a single beat estimation of M_w, V_w was approximated to V_0 [12], the x-axis intercept of the end systolic pressure volume relationship (ESPVR), which was estimated using a single beat estimation method for maximal elastance (E_{max}) based on the bilinear approximation of the ventricular elastance [13], [14]. This approximation was validated using data recorded during manual aortic clamp, which continuously changed the ventricular load. The analysis of the sensitivity of M_w to an eventual error in the estimation of V_w due to this approximation was based on $\frac{\partial M_w}{\partial V_w}$.

III. RESULTS

In other to use M_w for the assessment of cardiac function during assistance with rotary blood pumps, it is important to analyze the influence of the pump over three variables: SW, LVV_{ED} and V_w.

Fig. 2 shows changes of SW and pump external work (EW_{pump}) against changes in the N in the steady state data from Animal # 1.

The influence of N over LVV_{ED} is exemplified in Fig. 3, where are represented mean LVV_{ED} and mean LVV_{ES} measured at each N with Animal # 1.

Fig. 4 shows the relationship between V_w and V_0 when both where estimated using conventional multiple beat methods for data from all experiments. There was a significant ($p < 0.001$) correlation between V_w and V_0 (r=0.64).

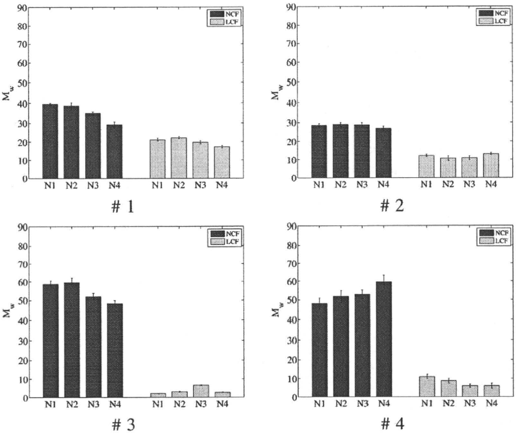


Fig. 5. Slope M_w estimated for 4 different mean pump rotational speed ($N1$-$N4$) at each cardiac function (NCF and LCF)

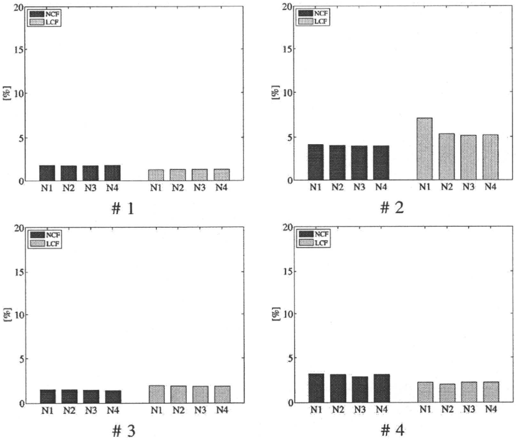


Fig. 6. $\frac{\partial M_w}{\partial V_w}$ as a percentage of M_w against pump rotational speed ($N1$-$N4$) for each cardiac function (NCF and LCF)

Fig. 5 represents the estimated slope M_w comparing data from NCF (dark-grey bars) and LCF (light-grey bars) at different rotational speeds.

Finally, Fig. 6 shows the sensitivity of M_w to V_0 as the percentage of M_w that $\partial M_w/\partial V_w$ represents.

IV. DISCUSSION

The preload recruitable stroke work is an important cardiac function index since its sensitivity to ventricular contractility is independent of ventricular afterload. Another important advantage is that the estimation of M_w does not require continuous assessment of LVV, which would require an invasive measurement.

Understanding the influence of the pump rotational speed over external work, represented in Fig. 2, is fundamental for the interpretation of M_w as a cardiac function index during the assistance. Different from pulsatile pumps, centrifugal pumps generate flow without directly generating pressure difference, therefore, as shown in Fig. 2, an increase of N resulted in an increase of the pump flow and, consequently, of EW_{pump}. At the same time, stroke volume decreases, decreasing SW. The increase of EW_{pump} and the decrease of SW were not equivalent, besides the relationship between those two external works was different for each animal and was also dependent on the cardiac function.

LVV_{ED} was expected to decrease as the diastolic pump flow increased when N was increased. However, LVV is directly influenced by physiological regulatory mechanisms and there was no constant relationship between LVV_{ED} and N in the data obtained in two levels of cardiac function and four experiments. Instead, changes in LVV_{ED} were similar to changes in SW, which indicates that LVV_{ED} unbiased by V_w could compensate changes in SW due to changes in pump rotational speed, maintaining M_w independently.

Next, for a single beat estimation we compared V_w and V_0 estimated as the x-intercept point of the linear regression between SW-LVV_{ED} and between LVP_{ES}-LVV_{ES}, respectively. The strong correlation between V_w and V_0 indicates that V_0 could be used also for the estimation of M_w. Methods for the estimation of the E_{max} and V_0 using single beat estimation methods during assistance with centrifugal rotary blood pumps have been analyzed in previous studies [14].

Finally, we compared M_w estimated for data recorded before and after the Propranolol injection. Although the estimated M_w varied with changes in N, there was no strong correlation between M_w and N. Small variations in the contractility are expected especially during open-chest measurements. M_w estimated for LCF was lower than M_w estimated for NCF in all data, which indicates that M_w was sensitive to change in the ventricular contractility.

Small errors in the estimation of V_w did not result in considerable changes in M_w. As shown in Fig. 6, $\frac{\partial M_w}{\partial V_w}$ was not higher than 10 % than M_w in all evaluated data. Therefore, inaccuracies in the approximation of V_w to V_0 do not affect the estimation considerably.

V. CONCLUSIONS AND FUTURE WORKS

Before the assessment of the myocardial recovery during the assistance with RBPs we should consider the influence of the pump rotational speed over the cardiac function index, as well as verify the sensitivity of the index to changes in the ventricular contractility.

In this paper we presented an evaluation of the influence of the assistance with a centrifugal rotary blood pump over the PRSW. Pump rotational speed influences the SW of the native heart; however, LVV_{ED} unbiased by V_w was also affected and could compensate this dependency. In the data set evaluated, M_w could detect changes in the cardiac function induced by Propranolol infusion.

This study has some limitations, such as the small data set that was analyzed. Future studies should also include other types of pump and take into account the differences between each pump. Moreover, in the present study, data of unassisted hearts were not evaluated, which is important for the validation of cardiac function indices during assistance with RBPs.

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Dynamic Characteristics between the Subjective Score of Motion Sickness Discomfort and Video Global Motion

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***Abstract*—It is well-known that visually-induced motion sickness (VIMS) is caused by image motion. Therefore it is important to clarify the relationship between image motion and the change in discomfort level. However, it is difficult to know the quick change in the level of discomfort during watching actual video image. The authors have proposed a method of interpolation for the subjective score, which has low time and quantitative resolutions, by using physiological parameters. The model which represents the change in subjective score of VIMS was expressed as multiple regression equations in which input parameters are cardiovascular indices such as heart rate variability. In this study, the model which represents the relation between global motion vectors of a video image and estimated subjective score was identified as ARX model. The results indicated that the simple ARX model can estimate the change in subjective score from global motion vectors.**

I. Introduction

VISUALLY-INDUCED motion sickness (VIMS) is a kind of motion sickness that occurs while the subject is watching a moving image displayed on a wide field display or screen[1]-[3]. In order to reduce the risks for VIMS, it is important to investigate the relationship between VIMS and each component of video motion such as transition and roll. Ujike *et al.* reported the effects of global motion which is consisted by roll, pitch and yaw [4]. However, actual video image consists of various components of motion and each motion component may have synergistic effect. Furthermore, it is difficult to evaluate the symptoms of VIMS continuously while watching video because the degree of symptoms is usually evaluated by subjective a score stored with intervals. Sugita *et al.* investigated the effect of visual stimulation by using ρ_{max}, a physiological index defined as the maximum cross-correlation coefficient between heart rate and blood pressure [5]. Our group has reported that a model consisting of physiological indices can effectively represent VIMS with higher time and quantitative resolution than a subjective score [6]. The aim of this study is to identify the relationship between global motion vectors of video and estimated subjective score of discomfort level using system identification method.

This study was subsidized by JKA through its Promotion funds from KEIRIN RACE and was supported by the Mechanical Social Systems Foundation and the Ministry of Economy, Trade and Industry.

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II. Methods

A. Experiments

A total of 86 subjects (26.4±8.1years) participated in the study. We used two video images. One consisted of 5.5min gray still image, 20min video image including various motion and 2min gray still image (movie A) (N=41) and the other consisted of same scenes in a different order (movie B) (N=45). The subjects watched the video image (movie A or B) displayed by a LC display (resolution 1920x1080). The size of the monitor was 37inches and viewing distance was 0.70m.

Electrocardiogram (ECG) and plethysmogram were recorded while watching the video. The subjective score (SS) of 4-level-graded (0 to 3) discomfort level was also recorded every 1min. After the trial, each participant filled out the Simulator Sickness Questionnaire (SSQ) [7].

B. Estimation of Subjective Score

The estimation model was represented by multiple regression equations as follows,

$$SS(k)=\sum_{i=1}^{7} p_i u_i(k)+q_0 \tag{1}$$

where k denotes the discrete time and u_i are 7 kinds of physiological indices which are mean heart rate (HR), mean pulse transit time (PTT), coefficient of variation of R-R intervals ($CVRR$), low-frequency power of heart rate variability (LF), high- frequency of that (HF), LF/HF ratio and ρ_{max} calculated with a time window of 1min shifted every 10s. p_i are coefficients and q_0 is a bias term.

C. System Identification between global motion vectors and Estimated Subjective Score

In this study, the model which represents the relation between global motion vectors of a video image and estimated subjective score was identified as following ARX model,

$$SS(k)=\sum_{l=1}^{L} a_l SS(k-l)+\sum_{i=1}^{3}\sum_{j=1}^{M} b_{ij} GM_i(k-j)+w(k) \tag{2}$$

where, L and M are orders of the model, GM_i are global motion vectors representing span, tilt and roll, a_i and b_i are coefficients and $w(k)$ is the residue assumed to be a white noise.

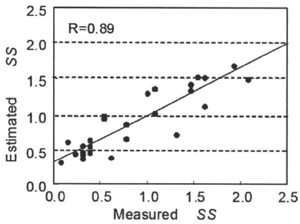

Fig. 1 Estimation result of subjective score (movie A).

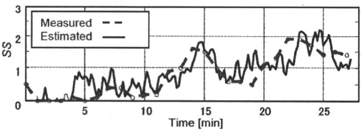

Fig. 2 *SS* estimation result by input physiological indices calculated every 10s (movie A).

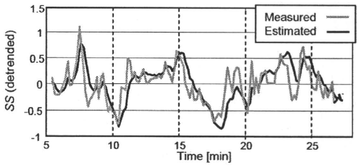

Fig. 3 *SS* estimation result by global motion vector (movie A).

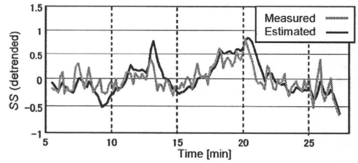

Fig. 4 *SS* estimation result by global motion vector of movie B using the same model as Fig.3.

III. Results and Discussions

Eleven (movie A) and 20 (movie B) subjects whose increment of the score of nausea in SSQ was over 25 were available for analysis. Fig.1 shows the mean result of *SS* estimation using multiple regression equations. The correlation coefficients were 0.89 for movie A and 0.88 for movie B. These results indicated that the change in *SS* can be estimated from physiological indices. Fig.2 shows the estimation result by using physiological data set which was calculated every 10s. This curve may represent continuous change in degree of VIMS.

Fig.3 shows the estimation result for mean change in continuous subjective score of which a first-order trend was removed. L and M in (2) were 1 and correlation coefficient was 0.74. Fig.4 shows the comparison of the change in continuous subjective score in movie B and the output of the model which was identified using the result of movie A with the global motion vector of movie B as input. The correlation coefficient was 0.80. This result may indicate that the simple ARX model can estimate the change in degree of VIMS from grovel motion vectors of a video image. However, the model could not estimate rapid change of the continuous subjective score when a first-order trend was not removed. It seems that the structure of the model which represents rapid change is different from the one that represents slow accumulable change.

IV. Conclusion

In order to evaluate the effect of VIMS induced by actual motion image which includes various and complex motion components, the system identification of dynamic characteristics between subjective score and video global motion was performed. One of the features of proposed method is to estimate the continuous change in a subjective score using physiological indices. The results indicate that simple ARX model can effectively represent the change in degree of VIMS which is estimated by multiple regression equations using physiological indices. On the other hand, the model could not estimate the accumulate change on the degree of VIMS. Therefore, it is necessary to investigate other kind of model such as non-linear model.

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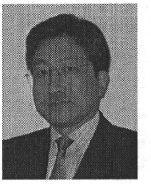
Assessment of Effects of Habitual Exercise on the Autonomic Nervous Function Using Plethysmogram



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Abstract

This article has proposed two methods for obtaining the linear correlation of the baroreflex system without measurement of blood pressure. One is based on the pulse wave transit time which needs both the electrocardiogram and the photoplethysmogram. The other is based on the photoplethysmogram only. The results from the experiments showed that the effect of habitual exercise and the Valsalva maneuver can be verified quantitatively. The proposed methods are possible to be used for a test of the autonomic nervous function at home.

1. Introduction

Japan is being a super-aging society, and thus the inflation of nationwide medical cost can be predicted exactly in the very near future. To prevent the crisis in the nation's deficit-ridden health insurance system, we should seriously consider some strategies for protecting people's health.

In this situation, one of the most effective methods is to entrench people to exercise habitually, which may be valid especially for people suffering from metabolic syndrome instructed to the specific medical checkup. To promote habitual exercise, some indices representing the effect of exercise should be feedback to the person after the exercise.

It has been indicated that the index corresponding to linear correlation of the baroreflex system is useful for expressing the autonomic nervous function [1,2]. We adopted this index as information to be fedback to the person exercising to settle it a habit.

Unfortunately this index needs measurement of continuous blood pressure to calculate. However it is not easy to measure continuous blood pressure because its sensor is too expensive and bulky to use at an ordinary home.

In this article, two alternative methods without measuring continuous blood pressure are introduced. One is a method in which the pulse wave transit time (PTT [ms]) is used instead of continuous blood pressure, and PTT is obtained from electrical cardiogram (ECG) and photoplethysmogram (PPG). The ECG is also difficult to measure at home and should not be used if possible. The other method does not need the ECG signal but linear correlation can be still obtained on the basis of only PPG signal.

In the use of these methods, the effect of habitual exercise on the index was evaluated and Valsalva maneuver was employed to ascertain the capability of the proposed method to extract individual difference from the calculated index.

2. Methods

2.1 Method Based on Pulse Wave Transit Time

2.1.1 Monitoring device

For home use, a measurement device should be wireless with less constraint. There are many monitoring systems but RF-ECG (Micro Medical Device, Inc.) is unique as a very small (40mm×35mm×7.4mm) and light (11.8g including electrical cell) wireless monitoring sensor to measure not only ECG but also acceleration and temperature as shown in Fig.1a).

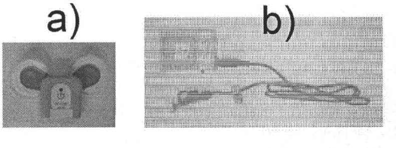

Fig. 1 Wireless sensors for measuring a)ECG signal and b) photo-plethysmographic signal

However, there is no small sized wireless device which can measure both ECG and PPG signals. In this study, we developed a sensor based on RF-ECG whose input terminal can acquire the PPG signal by attenuating its voltage level as shown in Fig.1b). To avoid the effect of body motion, the PPG signal was measured at the ear lobe and the sensor itself was inserted into a pocket on the chest of the subject's cloth. Another RF-ECG was used simultaneously to measure the ECG signal. Thus, both ECG and PPG signals can be measured in a wireless fashion at the sampling rate of 204Hz with less restriction.

The heart rate HR [bpm] was obtained from the reciprocal of the ECG signal, and PTT was calculated as the interval from the peak time of R-wave of the ECG signal to the peak time of the velocity of the PPG signal. Both PTT and HR were band-pass filtered with a pass band between 0.08Hz and 0.12Hz to be limited to the Mayer wave-related frequency components. After the processing, cross-correlation coefficient $\rho(\tau)$ between these signals was calculated time-discretely as follows:

$$\rho(\tau)=\frac{\phi_{PTT,HR}(\tau)}{\sqrt{\phi_{PTT,PTT}(0)\cdot\phi_{HR,HR}(0)}} \tag{1}$$

where $\phi_{PTT,HR}(\tau)$ is the cross-correlation function between PTT and HR, and $\phi_{PTT,PTT}(\tau)$ and $\phi_{HR,HR}(\tau)$ are auto-correlation functions of PTT and HR, respectively. In this study, $\rho(3)$, i.e., the value of $\rho(\tau)$ at $\tau=3$ s was obtained as an index which represents the linear correlation of the baroreflex system from PTT to HR. The index $\rho(3)$ is more stable value than the conventional index $\rho_{\max}$ which is defined as the maximum value of $\rho(\tau)$ in spite of its lower value.

2.1.2 Experiment

In the experiment, elderly people were used as test subjects classified into two groups. One is the Exercise Group consisting of 8 healthy people (age 52 to 73; mean 65.6±7.7; 4 males and 4 females) exercising habitually for over 15min a week. The other is the Control Group consisting of 8 almost healthy people (age 55 to 75; mean 67.6±6.5; 3 males and 5 females) exercising little

To give dynamic change in subject's hemodynamics by using change in his or her posture, the experimental protocol was as follows:

1) Supine position (5 min)
2) Upright standing position (2 min)
3) Supine position (3 min)
4) Upright standing position (2 min)
5) Supine position (3 min)

2.2 Method Based Only on Photoplethysmogram

2.2.1 Estimation of cross-correlation function

So far, many methods for obtaining blood pressure based on photoplethysmogram have been proposed, in which, for example, local maximum or minimum values of the acceleration of the signal are utilized. The purpose of these methods is usually to obtain the absolute value of blood pressure. However, if the purpose is to estimate the linear correlation of the baroreflex system, we can do it as shown below.

First, as shown in Fig.2, obtain the feature variables specifying the PPG signal at a certain beat such as the first extremum a and the second extremum b of the acceleration, the second extremum B of the velocity, the mean value MP and the difference PA between the maximum and minimum values of the signal within the beat. Since these variables are sampled every unequally-spaced interval, resample it every equally-spaced interval of 0.2s (5Hz) after the cubic spline interpolation. Let k be a discrete time which is incremented with the resampling, and produce a feature vector given by

$$x(k)=\left[a,b,b/a,a^2,b^2,ab,B/a,MP,PA\right]^T \tag{2}$$

Consider a multiple regression model in which an explanatory variable is $x(k)$ and an objective variable is heart rate $y(k)$ as follows:

$$\hat{y}(k)=\beta^T x(k)+\varepsilon(k) \tag{3}$$

where β is a coefficient vector to be identified with the least square method and $\varepsilon(k)$ is a residue. In general, it is expected that cross-correlation between blood pressure and heart rate whose frequency components are limited to the Mayer wave-related band is maximized a few second later. This phenomenon means that the baroreflex system has a delay. Let denote the delay as L [ms]. Unfortunately, the value of L is changed with time and subjects. Thus, find the optimal values β^* and L^* corresponding to β and L, respectively, so that the error between the $\hat{y}(k)$ and $y(k)$ can be minimized.

Instead of the value of (1), calculate a surrogate value that is obtained by letting $PTT=\hat{y}(k)$ and $HR=y(k)$ in (1). In this case, heart rate $HR=y(k)$ is calculated from the foot-to-foot interval (FFI [ms]) of the PPG.

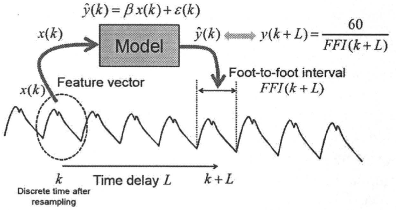

Fig.2 Multiple regression model with the input feature vector of pulse wave for estimation of heart rate.

The reason why the cross-correlation between heart rate and its estimate can be substituted by that between blood pressure and heart rate is shown below. It is known that heart rate correlates closely with blood pressure in the Mayer wave-rerated band at a resting state. If the output of the multiple regression model agrees well with the actual blood pressure, it is likely that the estimate obtained from the model also correlates well with heart rate.

On the other hand, the other method for estimation of blood pressure based on the PPG needs same calibration process using a blood pressure sensor. However, the proposed method employs the subject's heart rate as a reference value to identify the model parameters and does not needs any blood pressure sensor. While the method described in 2.1 which uses the PTT requires the ECG sensor to specify the position of the R-wave, the proposed method uses only the PPG sensor which is cheap and expected to be widely spread.

2.2.2 Estimation of cross-correlation function

Thirty-two healthy subjects (Age 23.1±3.6; 24 males; 8 females) were used in an experiment including the Valsalva maneuver with a protocol as follows:

1) Rest (5 min)
2) Respiration cease (1 min)
3) Rest (3 min)
4) Respiration cease (1 min)
5) Rest (5 min)

A photo sensor and an amplifier for (BIOPAC; PPG100C) was used to measure the PPG signal at the index finger and blood pressure sensor (Finapres; Portapres) was used to measure continuous blood pressure at the middle finger.

3. Results and Discussion

3.1 Method Based on Pulse Wave Transit Time

Figure 3 shows a subject's data (Age 70; female; systolic/diastolic pressure=136mmHg/84mmHg) who was a member of the Exercise Group. The data are time courses of heart rate, HR, the normalized PTT to have zero mean and unit standard deviation, and the cross-correlation coefficient at $\tau = 3$, $\rho(3)$. At two parts of the upright standing position, HR increased like a rectangular shape, which corresponds to normal orthostatic baroreflex to regulate blood pressure. It is found that PTT also increased in the similar manner as HR but their shapes are not so similar. Decrease in $\rho(3)$ can be seen at the two parts around upright standing positions.

Figure 4 shows another subject's data (Age 72; female; systolic / diastolic pressure= 155mmHg / 102mmHg) who was a member of the Control Group. Unlike Fig.4, hear rate HR changed irregularly, which means the baroreflex did not work well. Moreover, the shape of PTT is very similar to that of HR. This fact may be caused by arterial sclerosis since she had light hypertension and she was taking depressor drug routinely. That is to say, the change in HR may be equal to that of PTT if blood vessel is like a lead pipe with less compliance. The absolute value of $\rho(3)$ was small and drifted around zero during the experiment, which implies that the regulation function manipulating HR based on blood pressure with 3 seconds delay was not well done by the autonomic nervous system.

Figure 5 shows the comparison of the mean value $\rho(3)$ between the Exercise and the Control Groups. It is found that the recovery speed of $\rho(3)$ of the Control Group was significantly lower than that of the Exercise Group in the interval between two upright standing positions. It can be guessed that this difference was caused by the effect of habitual exercise.

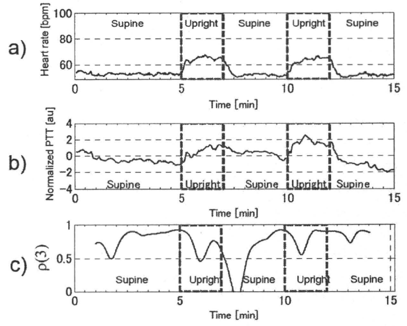

Fig.3 a)heart rate, b)normalized PTT and c)cross-correlation function $\rho(3)$ of a subject belonging to the Exercise Group (female; age 70; Systolic / diastolic pressure = 136mmHg/84mmHg).

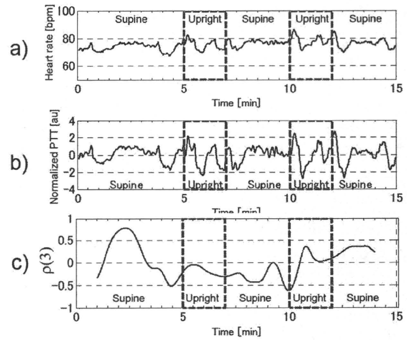

Fig.4 a)heart rate, b)normalized PTT and c)cross-correlation function $\rho(3)$ of another subject belonging to the Control Group (female; age 72; Systolic / diastolic pressure = 155mmHg /102mmHg).

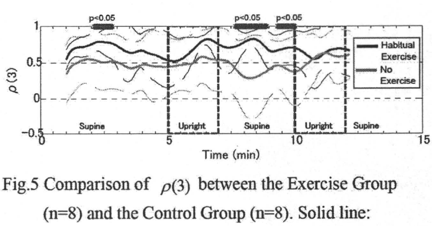

Fig.5 Comparison of $\rho(3)$ between the Exercise Group (n=8) and the Control Group (n=8). Solid line: mean. Broken line: S.D.

3.2 Method Based Only on Photoplethysmogram

The mean value of thirty-two subjects' $\rho(3)$ obtained from the method based only on the PPG signal described in 2.2 are shown in Fig.6, accompanied by $\rho(3)$ calculated directly from measured blood pressure. The figure indicates that $\rho(3)$ based on the PPG is much lower than $\rho(3)$ based on the blood pressure and that the effect of respiration cease on the time trajectories was not clear in both two kinds of $\rho(3)$. The reason is that the data shown in Fig.6 includes the subjects whose $\rho(3)$ is low even in the resting state. It can be guessed that such subjects tend to have a dull autonomic nervous reflex function against the respiration cease.

Thus, the subjects were selected under the condition that the mean value of $\rho(3)$ in the resting state from 0min to 4min was higher than 0.5. There were 9 subjects who satisfy the condition, and the mean value of $\rho(3)$ of these subjects is shown in Fig.7. It seems that the decrease in $\rho(3)$ around the positions of the respiration cease got deep.

In the same way, when the threshold to select the subjects was increased to 0.7, four subjects survived and their mean value of $\rho(3)$ was shown in Fig.8. It can be found that the depth of the decrease around the respiration cease tended to get deeper and that the value of $\rho(3)$ based on the PPG in the resting state came closer to that based on the blood pressure.

These results suggest that the temporal change in the linear correlation of the baroreflex system can be estimated on the basis of the PPG signal only without measurement of the ECG or blood pressure. Moreover, it is possible that $\rho(3)$ based on the PPG is more sensitive to the effect of the respiration cease than $\rho(3)$ based on blood pressure.

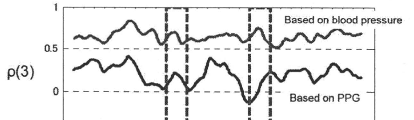

Fig. 6 Mean value of $\rho(3)$ of all 32 subjects.

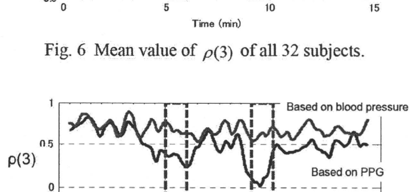

Fig.7 Mean value of $\rho(3)$ over 9 subjects whose mean value is over 0.5 from 0min to 4min.

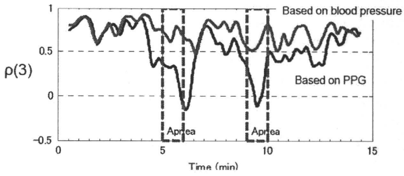

Fig.8 Mean value of $\rho(3)$ over 4 subjects whose mean value is over 0.7 from 0min to 4min.

4. Conclusions

This article has proposed two methods for obtaining the linear correlation of the baroreflex system without measurement of blood pressure. One is based on the pulse wave transit time which needs both the ECG signal and the photoplethysmogram. The other is based on the photoplethysmogram only. The results from the experiment with the former method indicated that the effect of habitual exercise can be verified quantitatively. The latter method suggested that the response of the linear correlation to the Valsalva maneuver can be estimated. The proposed methods are possible to be used for a test of the autonomic nervous function at home.

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脈波信号に基づく自律神経機能評価システムによる習慣的運動の効果判定

Assessment of Habitual Exercise Using the Autonomic Nervous Function Estimator Based on Plethysmogram

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Key Words: Autonomic nervous system, Habitual exercise, Plethysmogram

1. はじめに

少子超高齢化や医師の偏在などによる医療格差や地域医療の崩壊を防ぐための方法として，遠隔医療システムを利用した在宅療養や在宅リハビリテーションの実現が期待されている．

例えば，脳卒中後遺症患者は，退院後にも自宅において継続的にリハビリを行うことが望ましい．患者宅でもインターネット回線などを使うテレビ会議システムやSkypeなどのテレビ電話を利用すれば，病院にいる医師や作業療法士から遠隔的なリハビリ指導が受けられる可能性がある．

しかし,テレビ会議などを利用する遠隔的リハビリには，患者と指導者の間で交換できる情報が音声と映像だけである場合，的確な指導や患者動作の正確な把握が困難であるという問題がある．遠隔操作ができるロボットアームなどを使えば，作業療法士が実際に行うリハビリと同様に，患者の身体に触ったり動かしたりできる可能性がある．しかし，大掛かりで高価となり患者宅に広く普及する可能性は低い．これに対して，比較的安価なゴニオメータや加速度センサを使って患者の動作情報を計測・解析し，これを指導者に送ることができればリハビリの評価には有効であると思われる．

一方,従来の通常のリハビリ課題には単調なものが多く，遠隔的な環境で患者のリハビリへのモチベーションを保つことは容易ではない．これに対処するために，リハビリへのゲーム性の導入が以前から試みられてきた．この点，家庭用ゲーム機のコントローラに内蔵されている加速度センサ，あるいはボード型重心位置センサなどを使ったさまざまなゲームは，その内容を患者向けに変更することによってリハビリを楽しく継続でき，それをネットワークを通じて遠隔的に評価することができる可能性がある．

運動麻痺者に対するリハビリに限らず，特定健診対象者などに運動習慣を定着させるための方法としても，その効果を表す指標を提示することが考えられる．この指標として本研究では自律神経系が支配する圧反射特性の線形相関性を表す指標を採用した．ただし，家庭などでの応用を考えると血圧計測は困難である．そこで，血圧計測の代わりに心電図と脈波信号から得られる脈波伝搬時間を利用する方法，および，心電図も使わずに脈波信号だけから同様な指標を得る方法を開発した．ここでは，その指標によって習慣的運動の効果，および，バルサルバ試験による個人差の判定ができるかどうか検討した．

2. 方　法

2-1 概　要

非侵襲的な生体信号に基づいて自律神経機能を評価する伝統的な方法として，心電図や，計測が簡単な光電脈波センサに基づく心拍数情報を利用する方法がある．心拍数情報からは，周波数スペクトル解析やカオス・フラクタル解析などにより自律神経機能を推定できるとされている．

自律神経機能をモニタリングするための他の方法として，血圧変動と心拍数変動との間の関係である圧反射特性を求める方法がある．しかし，この方法では連続血圧の計測が必要であり，一般家庭では利用できない．

そこで本研究では，連続血圧の代わりに血圧情報を反映する脈波伝播時間の利用を考え，これを得るために脈波と心電図の両方を無線通信方式で計測する装置を開発した．また，心電図も使わずに脈波信号だけから同様な指標を得る方法を開発した．

2-2 脈波伝搬時間に基づく方法[1]

2-2-1 計測装置

心電図を無線通信を用いてモニタリングする装置はこれまでも多く存在するが，Fig,1a)に示す (株)医療電子科学研究所製のRF-ECGは，超小型（40mm×35mm×7.4mm）で軽量（電池を含めて 11.8g） であり無拘束性が高いばかりでなく，加速度・温度も同時に計測できる．一方，心電図と光電脈波の両方を同時に無線伝送する小型軽量の装置は存在しない．

そこで本研究では，Fig,1b)のようなRF-ECGの心電図入力端子に，振幅調整した光電脈波信号を入力するセンサを製作した．体動の影響をできるだけ避けるため，光ピックアップは耳朶上部に装着し，耳朶脈波を計測した．本体は衣服の胸ポケットに入れた．

これともう1台の心電図計測用のRF-ECGを同時に使うことにより，脈波と心電図の両方を無線通信方式により拘束性が低い状態のままサンプリング周波数204Hzで計測することができる．

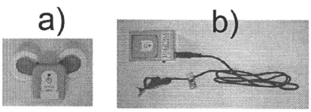

Fig. 1 Wireless sensors for measuring a)ECG signal and b) photo-plethysmographic signal

心電図信号から心拍数HR[min^{-1}]を求めるとともに，脈波伝搬時間PTT[ms]を，心電図のR波のピークの時刻から脈波信号の立ち上がり時刻までの時間として拍毎に求めた．PTT[ms]と心拍数HR[min^{-1}]は0.08Hz〜0.12Hzの帯域通過フィルタによってMayer波帯域に制限した後，PTTとHRの間の相互相関係数$\rho(\tau)$

$$\rho(\tau)=\frac{\phi_{PTT,HR}(\tau)}{\sqrt{\phi_{PTT,PTT}(0)\cdot\phi_{HR,HR}(0)}} \tag{1}$$

を時間離散的に計算した．ここで，$\phi_{PTT,HR}(\tau)$はPTTとHRとの間の相互相関関数である．また，$\phi_{PTT,PTT}(0)$および$\phi_{HR,HR}(0)$は，それぞれPTTおよびHRの自己相関関数である．$\rho(\tau)$の$\tau=3$[s]における値$\rho(3)$を圧反射系の線形相関性を代表する指標として求めた[2]．$\rho(3)$は，従来の$\rho(\tau)$の最大値$\rho_{\max}$に比べて，値は低いものの安定的に求められる指標である．

2-2-2 実 験

実験では，健康で1週間に1日以上1回15分間以上習慣的に運動をしている高齢者8名(**運動群**：52歳〜73歳；平均65.6±7.7歳；男性4名・女性4名)と日常的にほとんど運動をしていない高齢者8名(**非運動群**：55歳〜75歳；平均67.6±6.5歳；男性3名・女性5名)を対象として，次のプロトコル：

①仰臥位(5分)→②立位(2分)→③仰臥位(3分)→
④立位(2分)→⑤仰臥位(3分)

で仰臥位と立位の間の姿勢変化を与えた．

2-3 脈波信号のみに基づく方法

2-3-1 相互相関関数の推定方法

これまで，信号の加速度の極値を求めるなど，脈波信号を処理することによって血圧を推定する方法が多数提案されている．これらの方法では血圧の絶対値を推定することを目的としているが，圧反射系の線形相関性だけを求めるのであれば，以下に述べるような方法でこれを推定することが可能である．

まず安静時において，Fig.2のように，ある拍の脈波信号波形の特徴量として，a：加速度の第1極値，b：加速度の第2極値，B：速度の第2極値，MP：平均値，PA：拍内最大値と最小値の差を求める．これらの特徴量は拍毎の不等間隔で得られているので，拍間をスプライン関数で補間した後，0.2秒（5Hz）で再サンプリングする．再サンプリング毎に増える離散時間をkとしたとき，特徴量から成るベクトル

$$x(k)=\left[a,b,b/a,a^2,b^2,ab,B/a,MP,PA\right]^T \tag{2}$$

を説明変数とし，心拍数$y(k)$を目的変数とする重回帰モデルを

$$\hat{y}(k)=\beta^T x(k)+\varepsilon(k) \tag{3}$$

で表す．ここでβは重回帰係数ベクトルであり，$\varepsilon(k)$は残差である．βは最小2乗法で計算できる．通常，安静時のMayer波帯域における血圧から心拍数までの相互相関は数秒遅れて高くなることがわかっている．これは圧反射系に遅れが存在することに対応する．そこで，Lの値を変化させ，Lだけ先の心拍数の真値$y(k+L)$と推定値$\hat{y}(k)$の間の誤差が最も小さくなるような重回帰係数ベクトルβ^*と遅れL^*を決める．次に，(1)において$PTT=\hat{y}(k)$，$HR=y(k)$とおき，このモデルから得られる心拍数の推定値$\hat{y}(k)$とその真値$y(k)$との間のMayer波帯域における相互相関係数$\rho(\tau)$を計算し，これを血圧と心拍数の間の相互相関関数の代用とする．この場合，心拍数の真値$y(k)$は脈波信号の立ち上がり時刻の間隔FFIから求める．

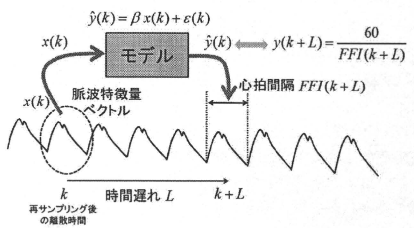

Fig.2 Multiple regression model with the input feature vector of pulse wave for estimation of heart rate.

脈波情報に基づく心拍数の推定値と真値との間の Mayer 波帯域における相互相関係数 $\rho(\tau)$ が，実測した血圧と心拍数の間のそれの代用とできる理由は次のとおりである．すなわち，通常，安静時の Mayer 波帯域における血圧と心拍数の間の相互相関は比較的高い．このため，脈波情報を入力とする重回帰モデルの出力が実際に血圧情報に近いものであるならば，この推定出力と心拍数の間の相互相関も当然高くなるはずだからである．

脈波の特徴量から血圧を推定する方法では，別な血圧計で推定モデルのパラメータの校正が必要となる．これに対し本方法では，対象者自身の心拍数を真値とするので，安静時に重回帰モデルを同定するだけでよい．また，脈波伝搬時間 *PTT* を使う方法では R 波の時刻を求めるために心電図の計測が必要であったのに対し，この方法では心電図計測が不要であり，安価な光電脈波センサだけを使って容易に $\rho(\tau)$ を推定できる．これにより，ゲーム機用脈波センサなどを使えば家庭でも容易に $\rho(\tau)$ が計測できるようになると思われる．

2-3-2　実　験

32 名の健常な被験者（23.1 歳±3.6，男 24 名，女 8 名）に対し，次のプロトコルの実験を行った．

①安静(5 分)→②呼吸停止(1 分)→③安静(3 分)→
④呼吸停止(1 分)→⑤安静(5 分)

座位において，脈波アンプ（BIOPAC 社 PPG100C)で右手人差し指における指尖脈波を計測し，連続血圧計（フィナプレス社 Portapres）で右手中指における連続血圧を計測した．

3. 結果および考察

3-1　脈波伝搬時間に基づく方法

Fig.3 は，運動群のある被験者(女性 70 歳；最高/最低血圧＝136mmHg/84mmHg)の心拍数 *HR*，平均 0・標準偏差 1 に規格化した脈波伝播時間 *PTT*，および相互相関係数 $\rho(3)$ の経時的変化を示している．2 箇所の立位時には起立性圧反射が生じ，*HR* が方形波状に増加していることがわかる．*PTT* も *HR* に伴って変化しているが，完全に相似的な変化をしているわけではない．$\rho(3)$ は第 1 回の立位に下降し，仰臥位では再び増加した．第 2 回目の立位では下降は浅かった．

Fig.4 は，非運動群のある被験者(女性 72 歳；最高/最低血圧＝155mmHg/102mmHg)の Fig.3 と同様な量を示したものである．Fig.3 と異なり，立位時には起立性圧反射が明確ではなく，不規則な変動をした．また，*PTT* は *HR* と非常によく似た変化を示している．この被験者は軽い高血圧症であり日常的に降圧剤を服用している．このため，動脈硬化により超低周波領域における *HR* の変化がそのまま *PTT* に現れている可能性がある．一方，$\rho(3)$ は実験中低く，0 レベルの回りを推移した．

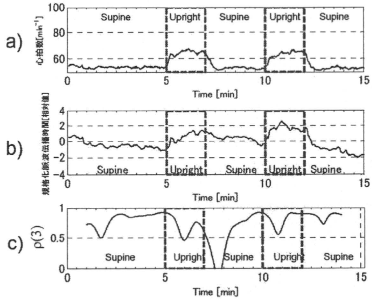

Fig.3　a)heart rate, b)normalized PTT and c)cross-correlation function ρ(3) of a subject belonging to the habitual exercise group (female age 70; Systolic/diastolic pressure=136mmHg/84mmHg).

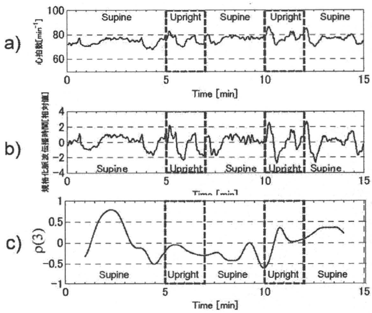

Fig. 4　a)heart rate, b)normalized PTT and c)cross-correlation function ρ(3) of another subject belonging to the no exercise group (female age 72; Systolic/diastolic pressure=155mmHg/102mmHg).

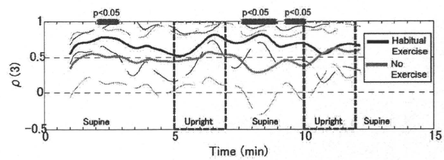

Fig.5 Comparison of between the habitual exercise group and the exercise group. Solid line: mean, Broken line: S.D.

Fig.5 は，運動群と非運動群の間の $\rho(3)$ の平均値を比較したものである．この図から，第1回目の立位から仰臥位に至る付近において，非運動群の $\rho(3)$ の回復が運動群より有意に遅いことがわかる．この差は日常的に運動を行っているかどうかで生まれた可能性が示唆される．

3-2 脈波信号のみに基づく方法

被験者32名全員について $\rho(3)$ を求めたものが Fig.6 である．図には同時に血圧と心拍数の間の $\rho(3)$ も示している．これらを見ると，脈波伝搬時間に基づく $\rho(3)$ の値は，血圧に基づくものよりかなり低い．また，どちらの $\rho(3)$ の変化も呼吸停止の影響がよくわからなかった．この理由は，被験者の中には安静時でも $\rho(3)$ が低いものが混じっており，このような被験者は呼吸停止時の自律神経反応が鈍いからである可能性がある．

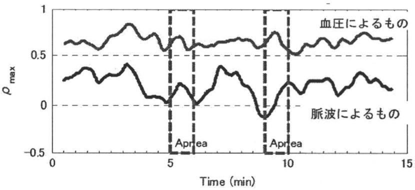

Fig. 6 Mean value of $\rho(3)$ of all 32 subjects.

そこで，推定心拍数に基づく0分から4分までの安静時の $\rho(3)$ の平均値が0.5以上の被験者だけを選ぶと9名おり，これらの被験者についてのみ $\rho(3)$ の平均値を求めると Fig.7 のようになった．この図から呼吸停止時の $\rho(3)$ の下降がやや大きくなったことがわかる．

同様にして，この閾値を0.6に上げると6名となり，Fig.8 のようになった．徐々に呼吸停止時の $\rho(3)$ の下降が著しくなっていることがわかる．しかも，安静時の推定心拍数に基づく $\rho(3)$ は血圧に基づく $\rho(3)$ に近接していくことがわかる．

以上から，心電図や血圧を計測することなしに脈波信号のみから圧反射系の線形相関性の経時的変化を把握できる可能性があることがわかる．しかも，血圧に基づく $\rho(3)$ より脈波に基づく $\rho(3)$ の方が，より顕著に呼吸停止の影響を表している．

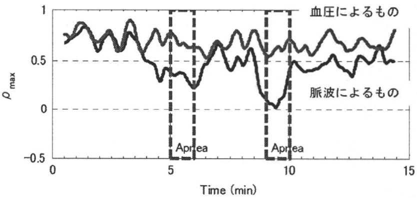

Fig.7 Mean value of $\rho(3)$ over 9 subjects whose mean value is over 0.5 from 0min to 4min.

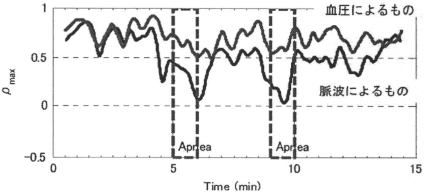

Fig.8 Mean value of $\rho(3)$ over 6 subjects whose mean value is over 0.6 from 0min to 4min.

4. おわりに

特定健診対象者などに運動習慣を定着させるための方法として，その効果を表す指標を提示することが考えられる．この指標として本研究では自律神経系が支配する圧反射特性の線形相関性を表す指標を採用した．ただし，家庭などでの応用を考えると血圧計測は困難である．そこで，血圧計測の代わりに心電図と脈波信号から得られる脈波伝搬時間を利用する方法，および，心電図も使わずに脈波信号だけから同様な指標を得る方法を開発した．その結果，習慣的運動の効果，および，バルサルバ試験による個人差の判定ができることが明らかとなった．

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総　説

診断治療機能を保持する自動車の可能性

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I. 緒　言

91 歳の高齢者が高速道路を逆走し，衝突した交通事故のニュースが報じられた．日本の法制上，高齢運転者の免許を剥奪する規定はないが，過去に類を見ない速度で高齢化社会を迎えつつある現在，高齢者の安全運転能力の問題は，社会経済学上も重要な課題になりつつある[1~3]．

夕張の例を語るまでもなく，地域を選ばず全国のいずれにおいても地方財政が完全に壊滅している日本，そして特に，インフラ整備の遅れた東北のような地方では，問題は一際，重要になることは論を待たない．公共の交通機関が壊滅している地域の高齢者から，乗用車を取り上げることなどできるはずもない．すなわち高齢者の安全運転の問題は，国民の基本的人権，生存権そのものにかかわる問題であり，この問題を放置することは，わが国においては憲法違反と言える．残念ながら単純に運動能力だけを比較すれば，若年者に比較すれば，高齢者は体力だけでなく，巧緻性その他においても，多少のハンディが存在する．自動車関連学会などでは統計上でも交通事故では高齢者の運転能力が問題に上がりつつあるのが現状である．様々な運転能力を比較する試験を行えば，高齢者は若年者と比較して，落ち着いて慌てず運転する長所はあるが，視力・有効視野の面ではハンディがあり，周辺運転状況の判断や，急な飛び出しなどに対する対応などには問題を残している[4,5]．

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（受領：平成 22 年 1 月 21 日）

憲法の精神による生存権の存続できる地域社会の安全な運転社会の構成のためには，すべての地域で公的移動機関の完璧なインフラ整備を行い，自動車の必要性をなくすることは解決策になりえる．しかしもちろん，東京一極集中により地域経済が自立できない政治の貧困を誇る現在では，完璧なインフラ整備実現の可能性はゼロである．もう一つの解決法は，自動車の完全な自動運転の具現化である．技術上は，そろそろ実現化の燭光も見えているが，社会的なインフラ整備やコスト，経済性を考えれば，まだ実用化は遥かに遠い．

高齢者安全運転社会実現のための三番目の解決策は，高齢者，あるいは肉体的ハンディキャップを持つ運転者において，残存する肉体能力を十全に発揮させ，あるいは，更に向上させることで，若年者に匹敵する安全運転．あるいは，それを若年者の運転の安全性を凌駕するような運転能力を完璧に発揮させることである．現実性を考えれば，唯一実現できる可能性がある選択肢と言える．そこで，東北大学モビリティ＆スマートエイジングプロジェクトでは，老化という自然現象に対して不自然に対抗しようとする抗加齢＝アンチエイジングではなく，高齢者の英知を大事にし，残された肉体能力を十二分以上に発揮させて，社会・経済・政治的な活動を，豊富な経験に則って，ますます頑張っていただこうという「スマートエイジング」の方向性を探求している[6]．

東北大学では，スマートエイジング社会実現を目指したプロジェクトの一環としてこの高齢者の健康増進，運転能力向上プログラムに着手し，様々な研究成果を上げつつあるので，本稿ではその一端を紹介し，高齢化社会における高齢者の乗用車運転の将来につい

て考察を加える．

II．健康を増進し，安全運転能力を向上させることができる乗用車の可能性

日本では18歳になれば乗用車の運転免許を取得することができる．若年ドライバーの運転の問題点として，特に追い越しなどの運転行動における行動特性において，安全運転上問題になるような乱暴な運転行動が多く，交通事故に結びつきやすい点などがある．これに対して，高齢ドライバーの問題点として，視力や視野角度などの側面において若年者と比較してハンディがあり，また急場における反応性においても，劣る面がある．長時間運転後の疲労蓄積の度合い，また疲労時の運転行動などにおいても，年齢で差がある側面は否定し切れないかもしれない[7,8]．

そこで，もし，高齢者が自家用車の運転中に，乗用車が運転者の自律神経の機能を適切に定量的に判定し，運転者の体調を整える何らかの治療的刺激を与え，自律神経の機能を安定して最適な運転行動を行い得る状態に維持することができれば，高齢者の安全運転に資することができる乗用車が具現化できることになる．

東北大学におけるモビリティ＆スマートエイジングプロジェクトは，この開発に着手し，様々な方向性を探っているが，そのコンセプトの具現化の1形態概念を図1に提示する[9]．

日本には自動車会社が比較的多いこともあり，乗用車を運転中の運転者の体調モニタリングには様々な方向性が試みられている．両手，あるいは両足，あるいは，心臓を挟んだ位置の体表面の電極からは，原理的には心電図記録が可能であり，図1に提示するような，ハンドルに装着した心電図以外にも，脈波モニター，ドライブシートを使ったセンサにも様々なシステムが各社で開発されている[10,11]．自動車関連企業の協力により加齢医学研究所ではハンドル装着型心電図モニター装置の開発に関しては，これまで長年従事してきた伝統があり，急な運転動作を行ったりしなければ，かなり安定した記録が可能であることは確認されている．また，脈波計測システムもハンドル装着型は比較的，実現の可能性は早いと考えている．光電式の容積圧脈波計測システムはシステムが安価で，汎用性が高く，どのような乗り物にも応用しやすいシステムである[12,13]．

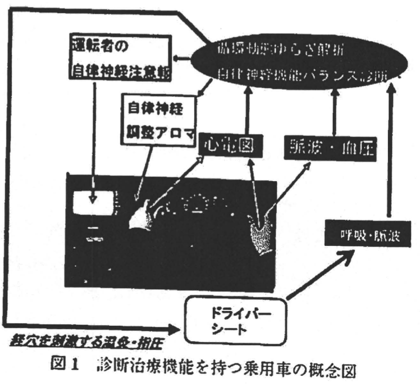

図1　診断治療機能を持つ乗用車の概念図

これまでに脈波を使うことで，血管年齢に代表される体調モニターを具現化できるという方向性は多くの研究者によって進められており[12~14]，脈波系の計測結果による時系列曲線を二回微分して加速度脈波として解析すれば，トレンド成分を除去することができ，反射波の影響を定量的に評価できる．このような方法論により，血管年齢指数に代表されるようなパラメータが計算できるので，原理的には，自動車のドライバーは，ハンドルを握って約1秒，すなわち1心拍分の記録時間を要するだけで，その日の体調が，カーナビディスプレイに表示できることになる．血管年齢指数の計算の方法論については，様々な施設において，様々な方法論が提案され，報告されている[14~18]．

東北大学で研究が進むシステムでは，ドライバーシートにエアバッグ式のセンサが内蔵される計画である．このシステムにより，運転者の呼吸及び脈波を計測することができる．東北大学で実験が進むドライビングシート内蔵型のエアバッグシステムを図2に提示する[20]．

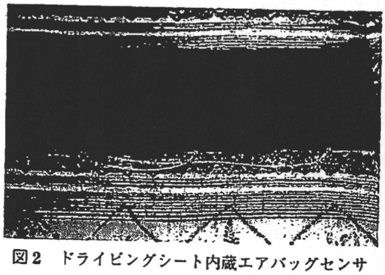
図2　ドライビングシート内蔵エアバッグセンサ

図3に提示するように，システムにおいては，エアバッグ時系列信号から信号処理の技術で，呼吸成分と脈波成分を抽出することが可能になっている．図4に提示するように呼吸信号が，また図5に提示するように，心拍動に一致する時系列信号が記録可能であり，この脈波成分の成因は，大動脈の拍動に起因することが明らかになっている．従って，この時系列信号にハンドル心電図の解析結果と合わせれば，脈波伝播速度の計算も可能になる．

脈波伝播速度は，健康な被験者では血圧に比例することも明らかになっており，腕時計型の血圧計などで商品化もされているので，原理的には，運転者がドライバーシートに座ってハンドルを握ったとたん，約1心拍前後の計測時間で，血圧の算出が可能になる．従って，洞調律であれば原理的には1秒前後で，心電図，血圧と並んで，加速度脈波による加速度脈波化例指数(SDPTGAI)，脈波伝播速度やCAVIに換算される**stiffness parameter** βから定量された血管年齢指数などで表される健康パラメータまで，自動車の始動時にカーナビのディスプレイに表示できることになる．

更に，図4に提示するように運転者の呼吸成分の抽出も可能である．運転のスタート時に，エンジンの暖気運転を行い，急発進をせずに，数回呼吸するまでの計測時間を待ってもらえれば，呼吸や心拍血圧のゆらぎまで解析して表示することができる．これにより，交感神経と副交感神経のバランスが解析できるので，運転手の状態に応じた適切な音声アドバイスを与えることも可能になる．現在，カーナビゲーションシステムなどに，音声アドバイス機構が設置されているが，道案内代わりに健康案内を行うことも，適切な定量診断の上であれば，可能であるかもしれない．

更に，慢性管理のために，ホルター心電図で解析されているような，呼吸成分や，低周波ゆらぎ成分，LF/HF成分のような解析も行うことができる．更に，心拍変動，脈圧変動，脈波伝播時間変動，血圧反動，そして呼吸変動などのマルチパラメータ計測が可能になるので，時系列信号の相互関係から，血圧反射機能等の算出も可能になる．東北大学では，心拍の反応性に現れる心臓血圧反射機能だけでなく，動脈の血圧反射機能も計算できる方法論を開発しているので，生体の多次元診断が具現化できることになる．これらのデータは，自動車のカーナビのパソコン内にストックすることもできる．また，セキュアなシステム開発を行えば，そのまま病院にインターネットでリンクすることも可能になるので，不整脈の発生時などの緊急アラートシステムとしても展開することが可能になる．

これらのデータに基づき，運転者の体調に応じて適切なアロマ刺激を空調で供給し，また，自律神経機能の状態に応じて，経穴に対する適切なマッサージ刺激や，温灸，冷灸などの東洋医学，あるいは，統合医療における治療原理に準じた適切な刺激を与えることで，運転者の状態を，健康な方向へシフトさせ，運転

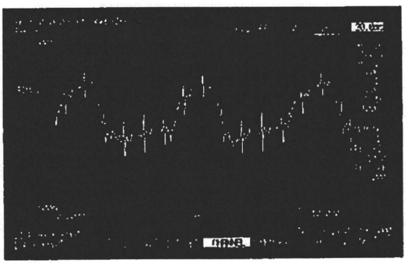

図3　エアバッグ空気室からの抽出信号

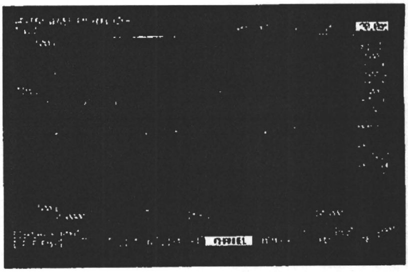

図4　抽出された呼吸信号

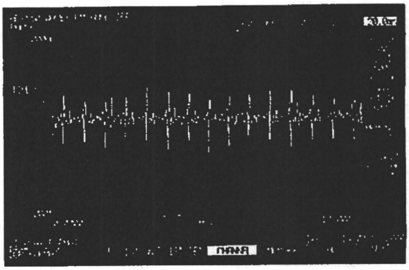

図5　抽出された拍動信号