E. 結論

電子診療鞄により被災者の自律神経機能障害が可能である。また、脈診機をもちいて脈診の客観的、定量的なエビデンスの構築が可能となることが示唆された。

図５．自律神経機能診断を行うことができる被災地における電子診療鞄

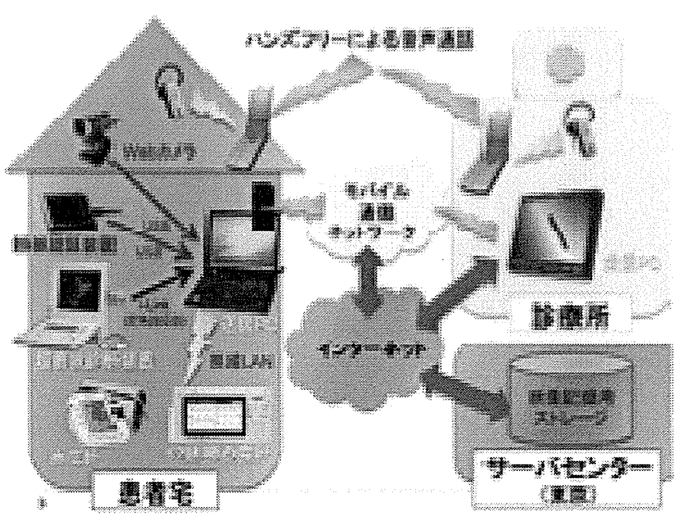

図６．電子診療鞄による医療支援

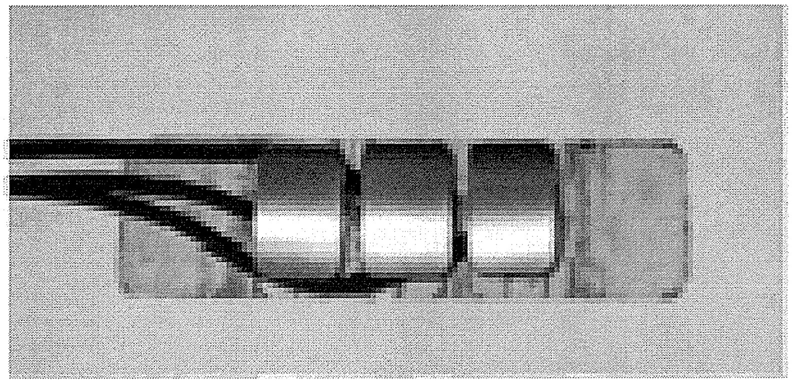

図７．脈診センサ

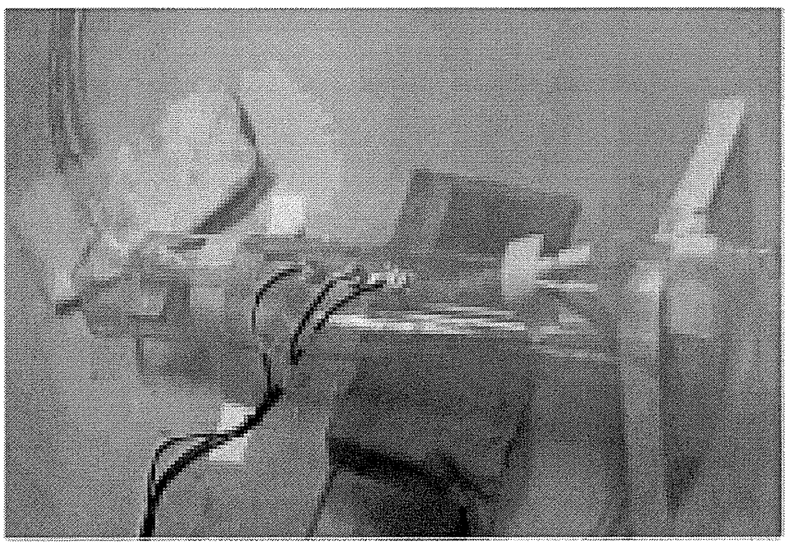

図８．モック循環回路におけるモデル橈骨動脈の脈診波形実験

平成 24 年度　統合医療が心臓血管機能に与える影響を解析するための
心臓内血流構造解析の研究

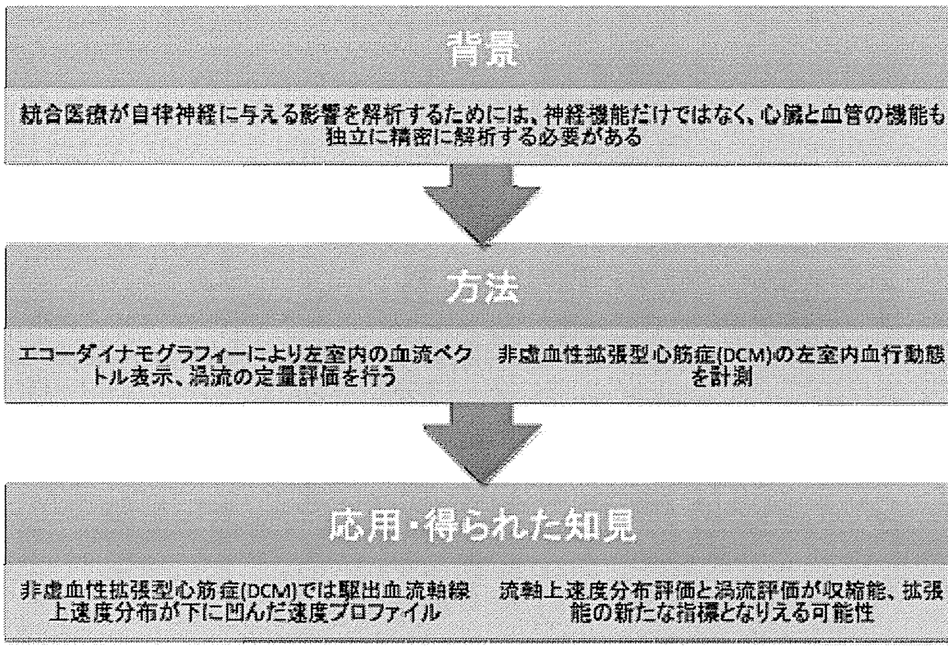

図 9．平成 24 年度の研究の全体像

A. 研究目的
統合医療が自律神経に与える影響を解析するためには、神経機能だけではなく、心臓と血管の機能も独立に精密に解析する必要がある。エコーダイナモグラフィーを用いて、非虚血性拡張型心筋症(DCM) の左室内血行動態を計測し、左室内渦流の各時相における渦の大きさについて定量計測を行い、収縮-拡張指標との関係を明らかにする。

B. 研究方法
正常心機能 10 例と非虚血性 DCM13 例を対象に、左室内の血流構造を血流ベクトル、渦流量、渦径、流軸線上速度分布を用いて検討した。

C. 研究結果
収縮期を通して DCM 心では正常心より大きな渦流が認められ、また、駆出血流軸線上速度分布では、正常での直線状の速度プロファイルに対し、DCM では下に凹んだ速度プロファイルが認められた。

D. 考察
血流ベクトル、渦流量および渦径により、左室心尖部長軸断面で収縮期の正常心、DCM 心を比較すると、収縮期を通して DCM 心では大きな渦流が認められたが、DCM では拡大した左室と収縮力の低下、拡張能の低下が影響として考えられる。

E. 結論
エコーダイナモグラフィーにより拡張と収縮は、左室内血流動態という新たな評価指標により互いに連関していることが判明し、流軸上速度分布評価と渦流評価が収縮能、拡張能の新たな指標となりえる可能性が示された。

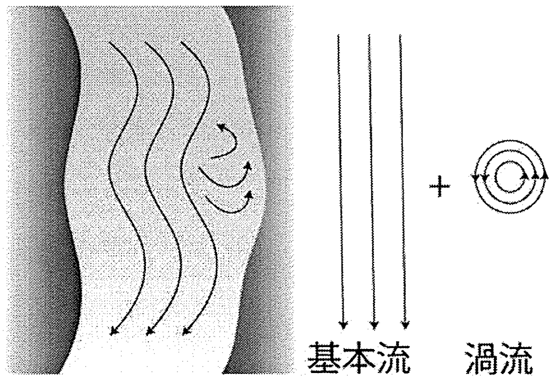

図 6

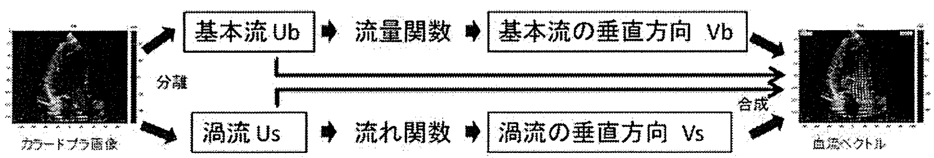

図 10. エコーダイナモグラフィ

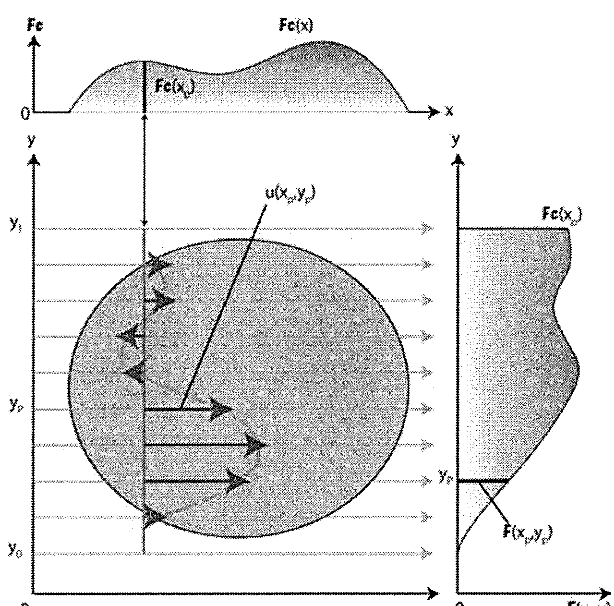

図 11. 二次元流れと三次元流れ

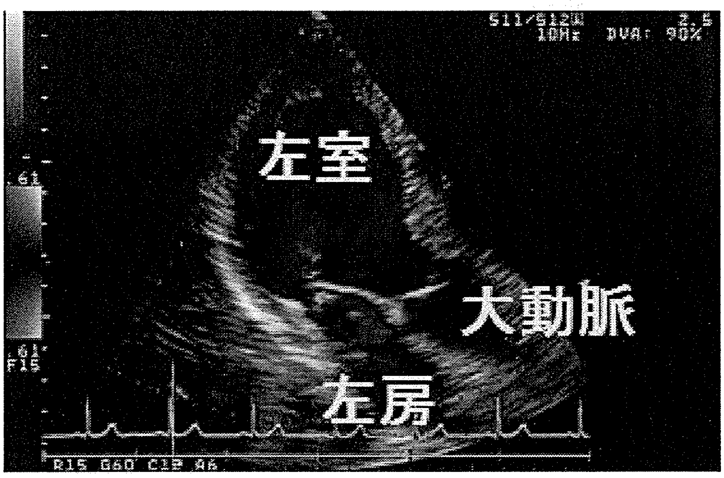

図 12. 心腔内血流カラードップラ表示

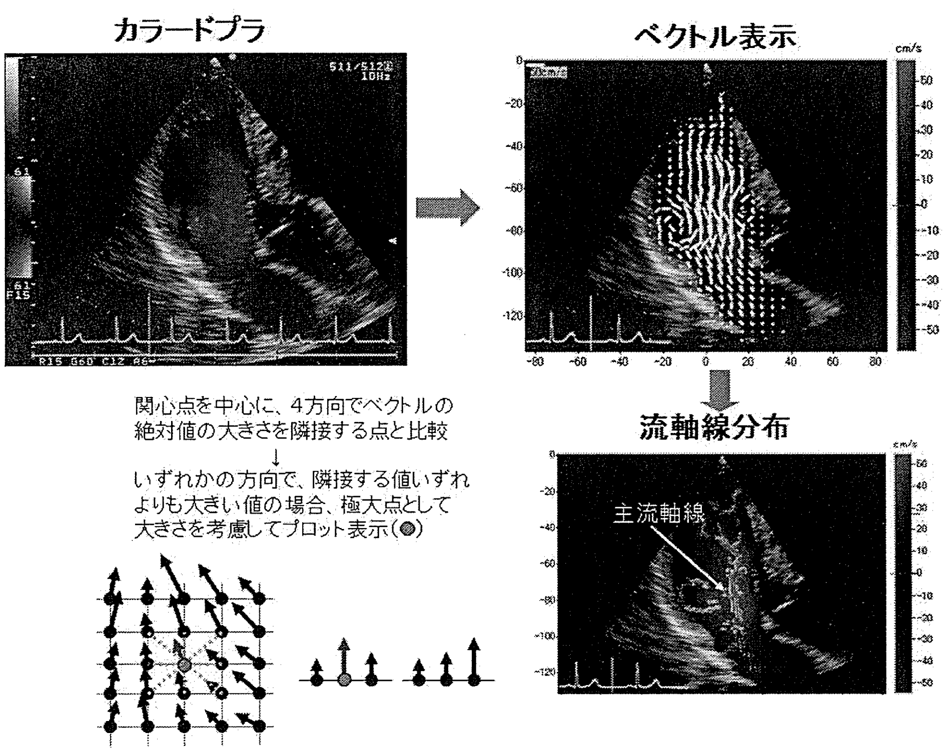

図 13. 主流軸上速度分布

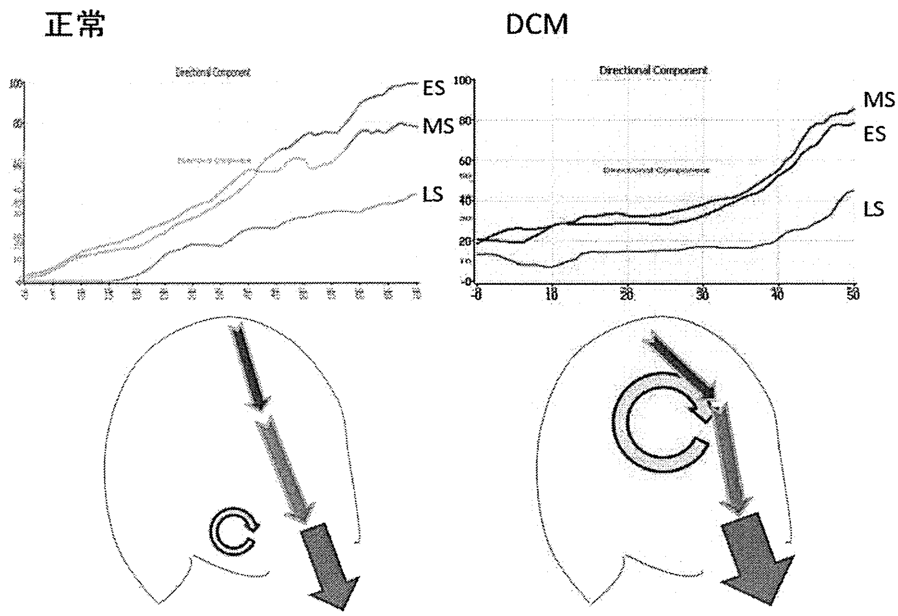

図 14. 流軸線上速度分布比較

Ⅲ.研究成果の刊行に関する一覧表

研究成果の刊行に関する一覧表

書籍

著者氏名	論文タイトル名	書籍全体の編集者名	書　籍　名	出版社名	出版地	出版年	ページ
Makoto Yoshizawa, Norihiro Sugita, Tomoyuki Yambe, Satoshi Konno, Telma Keiko Sugai, Makoto Abe, Noriyasu Homma, Shin-ichi Nitta	Methods for Estimating a Cross-Correlation Index of the Baroreflex System by Using a Plethysmogram	Takami Yamaguchi	Nano-Biomedical Engineering 2012	Imperial College Press	London	2012	566-576
Makoto Abe, Telma Keiko Sugai, Makoto Yoshizawa, Kazuo Shimizu, Moe Goto, Masashi Inagaki, Masaru Sugimachi, Kenji Sunagawa	Detection of Life-Thretning Arrhythmias Using Multiple regression Model	Takami Yamaguchi	Nano-Biomedical Engineering 2012	Imperial College Press	London	2012	577-586

雑誌

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阿部　誠, テルマ・ケイコ・スガイ, 吉澤　誠, 山家　智之, 清水一夫, 後藤　萌, 稲垣　正司, 杉町　勝, 砂川　賢二	重回帰分析を用いた致死性不整脈検出アルゴリズムに関する検討	生体医工学	48巻, 6号	577-583.	2010
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NorihiroSugita, MakotoYoshizawa, AkiraTanaka, Makoto Abe, Noriyasu Homma, Shigeru Chiba, Tomoyuki Yambe, Shin-ichi Nitta	Evaluation of temporal relationship between a physiological index and a subjective score using average mutual information	Displays	32	201-208	2011
阿部　誠，テルマ　ケイコ　スガイ，吉澤　誠，本間　経康，杉田　典大，清水　一夫，後藤　萌，稲垣　正司，杉町　勝，砂川　賢二	植込み型除細動器用致死性不整脈検出アルゴリズムの高速・高精度化	生体医工学	49(6)	932-938	2011
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:田中明(福島大学 共生システム理工学類)，杉田典大，吉澤誠，山家智之	「3D映像視聴による自律神経への影響 循環調節系の変化と3D映像視聴」	自律神経	48-3	211-213	2011

阿部　誠，吉澤　誠，テルマ　ケイコ　スガイ，本間　経康，杉田　典大，清水　一夫，後藤　萌，稲垣　正司，杉町　勝，砂川　賢二	植込み型除細動器への実装を考慮した致死性不整脈検出アルゴリズムの改良	電気学会論文誌C,	132(12)	1943-1948	2012
Makoto YOSHIZAWA, Tomoyuki YAMBE, Norihiro SUGITA, Satoshi KONNO, Makoto ABE, Noriyasu HOMMA, Futoshi TAKEI, Katsuhiko YOKOTA, Yoshifumi SAIJO, Shin-ichi NITTA	Application of a Telemedical Tool in an Isolated Island and a Disaster Area of the Great East Japan Earthquake	IEICE TRANSACTIONS on Communications	E95-B(10)	3067-3073	2012
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Ⅳ.研究成果の刊行物・別刷

Artif Life Robotics (2010) 15:522–525
DOI 10.1007/s10015-010-0861-0

ORIGINAL ARTICLE

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Extraction of the Mayer wave component in blood pressure from the instantaneous phase difference between electrocardiograms and photoplethysmograms

Received and accepted: August 11, 2010

Abstract To analyze the human baroreflex function noninvasively, a beat-by-beat blood pressure signal is often measured by tonometry or plethysmogram using continuous blood pressure sensors. However, these sensors are too expensive and bulky to be used for home healthcare or telemedicine. On the other hand, it is well known that the pulse transmission time (PTT) is strongly correlated with the beat-by-beat blood pressure, especially in the Mayer wave-related frequency band (0.05–0.15 Hz). To obtain a new physiological parameter with a higher correlation with blood pressure in this band compared with that obtained using the PTT alone, we proposed the same-phase temporal difference (SPTD) based on electrocardiogram and photoplethysmogram signals. By examining 94 healthy subjects using a combination of SPTD and conventional PTT, it was revealed that the correlation with blood pressure for 25 subjects could be improved using the SPTD instead of the PTT, by applying the criterion of the ratio of their powers. However, for 7 subjects, the correlation decreased.

Key words Blood pressure · Pulse transmission time · Instantaneous phase · Photoplethysmogram

1 Introduction

Escalating medical costs caused by rapid aging of the population, and health disparities caused by a shortage of physicians, are serious problems in Japan. Telemedicine could be a viable solution to these problems. For telemedicine to be functional, however, doctors require physiological information such as the electrocardiograms (ECG) and blood pressure of patients in remote places.

In this study, we focused on the baroreflex system as a new physiological information source for telemedicine. The baroreflex system is a negative feedback control mechanism in the autonomic nervous system that can attenuate the effects of perturbations in arterial blood pressure by changing the heart rate or vascular resistance; cardiovascular diseases such as hypertension are associated with malfunctions of this system.[1–4]

To estimate the baroreflex characteristics, a beat-by-beat blood pressure signal is frequently measured by tonometry or plethysmogram using continuous blood pressure sensors. However, these sensors are too expensive and bulky to be used for home healthcare or telemedicine. On the other hand, it is well known that the pulse transmission time, or pulse transit time (PTT), is strongly correlated with the beat-by-beat blood pressure,[5] especially in the so-called Mayer wave-related frequency band (0.05–0.15 Hz). To obtain a more accurate PTT signal in the Mayer wave-related band instead of blood pressure, this study proposes a new algorithm based on the instantaneous phase difference between ECG and photoplethysmograms (PPG).

As shown in Fig. 1, the PTT is defined as the time delay between the R-wave in the ECG and the pulse arrival at a peripheral point, for example, at a finger tip. Therefore, the accuracy of PTT depends on how these time points are detected. Most conventional methods use the moment when the finger PPG signal begins to rise after the R-wave appears as the arrival time. However, the arrival time obtained from a PPG is easily disturbed by noise and artifacts. There are several methods to detect the arrival time,[6] but it is not clear which is the most appropriate method for calculating the PTT. In this study, we propose a new method for calculating the PTT on the basis of the instantaneous phase difference between the ECG and PPG waves. The results are compared with those of the conventional method.

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This work was presented in part at the 15th International Symposium on Artificial Life and Robotics, Oita, Japan, February 4–6, 2010

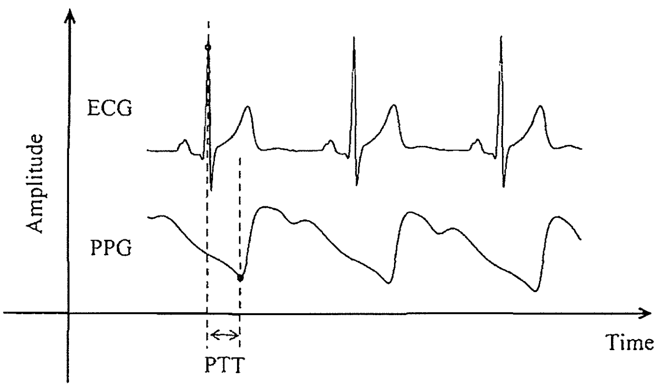

Fig. 1. Definition of pulse transmission time (PTT)

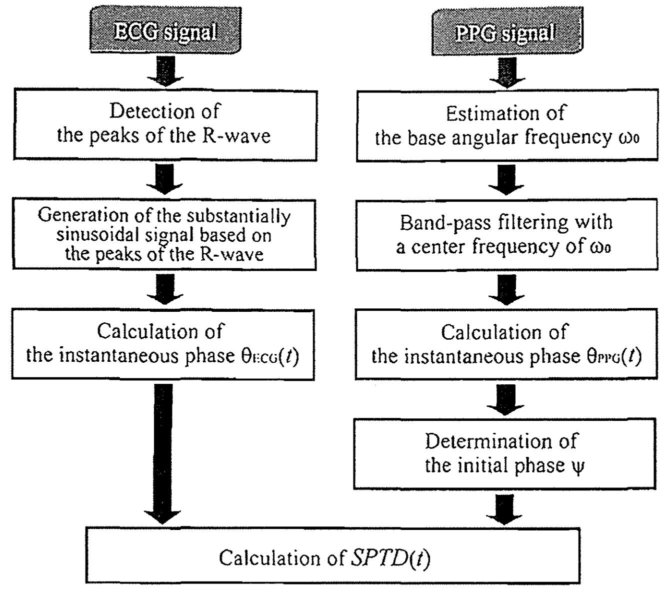

Fig. 2. Flowchart of the calculation of SPTD

2 Methods

2.1 Same-phase temporal difference

As a new index for determining the PTT, we propose the same-phase temporal difference (SPTD) between the ECG and PPG signals, as follows.

Suppose that a biological signal $s(t)$ at time t is given by

$$s(t) = \sum_k C_k \cos\theta_k(t) \tag{1}$$

$$\theta_k(t) = k\omega_0 t + \psi_k + \phi_k(t) \tag{2}$$

where C_k, Ψ_k, and $\phi_k(t)$ are a Fourier coefficient, an initial phase, and an instantaneous phase of the k-th harmonic wave, respectively, and ω_0 is a base angular frequency. If $k = 1$ and $t = t_m$, which is the time of the appearance of a feature point (e.g., the peak of an ECG R-wave) at the m-th beat, Eq. 2 is represented by

$$\theta_1(t_m) = \omega_0 t_m + \psi_1 + \phi_1(t_m) = 2m\pi \tag{3}$$

So the time interval TI_m between the feature point at the m-th beat and that at the $(m + 1)$-th beat is given by

$$TI_m = \theta_1^{-1}(2(m+1)\pi) - \theta_1^{-1}(2m\pi) \tag{4}$$

where $\theta_1^{-1}(\bullet)$ is the inverse function of $\theta_1(\bullet)$. Here, suppose that $s(t)$ is an ECG signal. Then TI_m indicates the R–R interval at the m-th beat. In the same way, if $s(t)$ is a PPG signal, TI_m is the heartbeat interval obtained from the PPG signal.

The PTT is considered to be the time interval between the feature point on the ECG and that on the PPG at the same beat. The SPTD at the m-th beat, $SPTD_m$, is defined as

$$SPTD_m = \theta_{ECG}^{-1}(2m\pi) - \theta_{PPG}^{-1}(2m\pi) \tag{5}$$

where $\theta_{ECG}^{-1}(2m\pi)$ and $\theta_{PPG}^{-1}(2m\pi)$ are the inverse functions of the corresponding phase function shown in Eq. 3 when $s(t)$ is the ECG and PPG, respectively. $SPTD_m$ can be regarded as the temporal difference between the fundamental harmonics of the ECG and PPG signals.

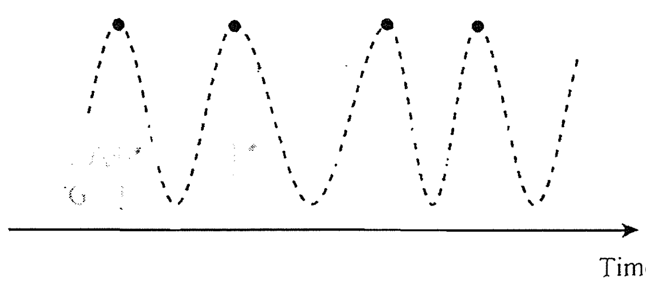

Fig. 3. Substantially sinusoidal signal whose crests appeared at the time points of R-wave peaks

2.2 Calculation method

Figure 2 shows a flowchart of the SPTD calculation. R-waves in the ECG were detected, and a substantially sinusoidal signal whose crests appeared at time points of the R-wave peaks was generated, as shown in Fig. 3. The base angular frequency ω_0 of the PPG was obtained as the peak frequency in the range 0.3–2.0 Hz. A 5th-order Butterworth band-pass filter with a center frequency of ω_0 and a bandwidth of 0.3 Hz was applied to the PPG signal. After these processes, the time series of the instantaneous phases $\theta_{ECG}(t)$ and $\theta_{PPG}(t)$ were calculated using the Hilbert transform. The initial phase of $\theta_{PPG}(t)$, $\Psi_{1,PPG}$, was set to minimize the difference in the mean value between the SPTD and the conventional PTT, in which the PTT was obtained as the delay between the peak of the R-wave and the rise time of the PPG. Finally, the time series of the SPTD was given by

$$SPTD(t) = \theta_{PPG}^{-1}[\theta_{ECG}(t)] - t \tag{6}$$

Furthermore, the maximum cross-correlation coefficient R_{max} between the PTT variability and blood pressure variability, whose frequency components were limited to the

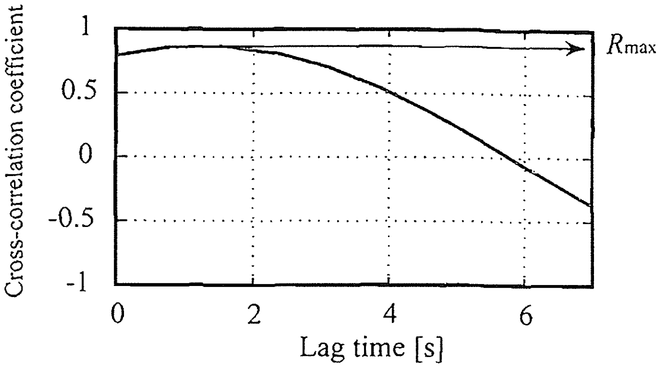

Fig. 4. Definition of the maximum cross-correlation coefficient R_{max} between two parameters

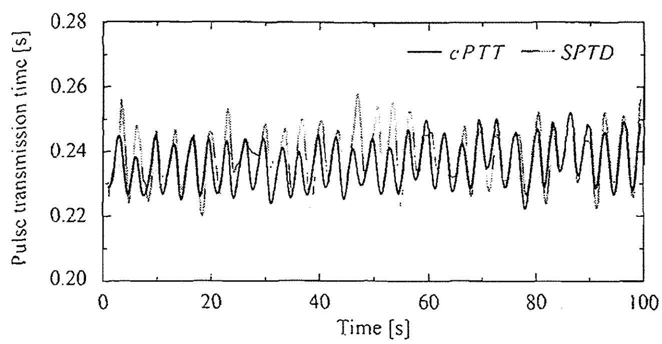

Fig. 5. Changes in the Mayer wave component of the *cPTT* and *SPTD* of a subject

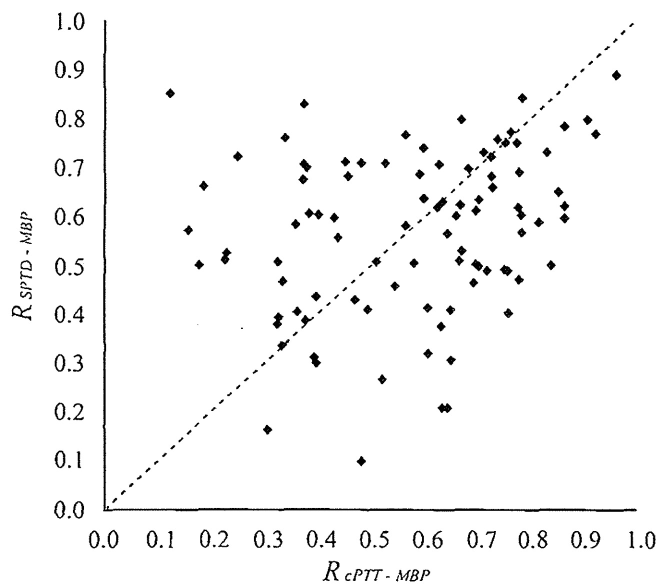

Fig. 6. Scatter diagram of $R_{cPTT\text{-}MBP}$ versus $R_{SPTD\text{-}MBP}$

Mayer wave-related band, was defined as shown in Fig. 4. It is possible to investigate the correlation between two physiological parameters in considering of their lags by introducing R_{max}. In this study, R_{max} between the conventional PTT (*-cPTT*) and the mean blood pressure (*MBP*), $R_{cPTT\text{-}MBP}$, and that between *-SPTD* and *MBP*, $R_{SPTD\text{-}MBP}$, were calculated. The minus sign was introduced as *-cPTT* and *-SPTD* because these parameters are negatively correlated with *MBP*.

2.3 Data acquisition

Physiological data were acquired from 94 healthy adults (69 males and 25 females, aged 23.4 ± 2.61 years). Informed consent was obtained from all subjects before the experiment. Each test subject was instructed to sit on a chair for 5 min, and their ECG and PPG were measured by electrodes placed on their chest and a photoplethysmographic sensor attached to a finger tip. The continuous arterial blood pressure signal was measured non-invasively using a finger blood pressure cuff (Portapres-Model 2; TNO-TPD Biomedical Instrumentation) or a tonometric pressure sensor (JENTOW 7700; Nihon Corin). These signals were amplified and converted to digital data by a 16-bit A/D converter (MP100; BIOPAC System). The sampling frequency was 1 kHz.

3 Results and discussion

Figure 5 shows changes in the Mayer wave component of a subject's *cPTT* and *SPTD*. As shown in this figure, these changes were very similar.

Figure 6 shows the scatter diagram for $R_{cPTT\text{-}MBP}$ and $R_{SPTD\text{-}MBP}$. In this figure, a dot above the diagonal dashed line represents a subject whose *SPTD* has a higher correlation with *MBP* than the *cPTT* does. This result indicates that the information of the *SPTD* is different from that of the *cPTT*, and that to get the highest possible correlation with blood pressure for some subjects, we should use *SPTD* instead of *cPTT*. To do this, we need a criterion for determining whether to adopt *SPTD* instead of *cPTT* without measuring the blood pressure. For this purpose, we introduced the power ratio (*PR*) of *SPTD* to *cPTT*. The index *PR* is defined as

$$PR = \sqrt{\frac{PW_{SPTD}}{PW_{cPTT}}} \tag{7}$$

where PW_{SPTD} and PW_{cPTT} are the powers of *SPTD* and *cPTT* in the Mayer wave-related band, respectively. Figure 7 shows the scatter diagram of the difference value between $R_{SPTD\text{-}MBP}$ and $R_{cPTT\text{-}MBP}$ versus the *PR*. This figure indicates that the difference between the two maximum cross-correlations R_{max}s increases as *PR* increases, and that $R_{SPTD\text{-}MBP}$ is higher than $R_{cPTT\text{-}MBP}$ in most subjects whose *PR* is larger than 3. Thus, a new index *nPTT* is defined as

$$nPTT = \begin{cases} SPTD & \text{if } PR \geq 3 \\ cPTT & \text{otherwise} \end{cases} \tag{8}$$

The scatter diagram of $R_{nPTT\text{-}MBP}$, which is the maximum cross-correlation between *-nPTT* and *MBP*, versus $R_{cPTT\text{-}MBP}$ is shown in Fig. 8. This result shows that using *SPTD* instead

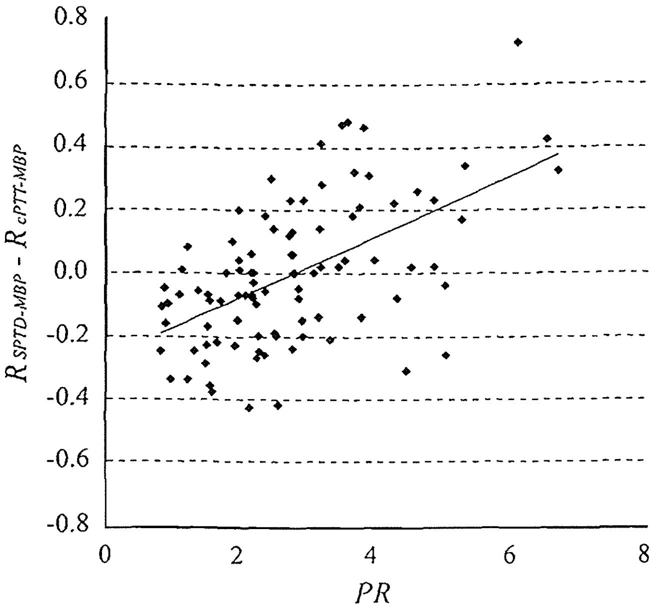

Fig. 7. Scatter diagram of the difference value between $R_{SPTD\text{-}MBP}$ and $R_{cPTT\text{-}MBP}$ versus PR. PR is the power ratio of $SPTD$ to $cPTT$ in the Mayer wave-related band

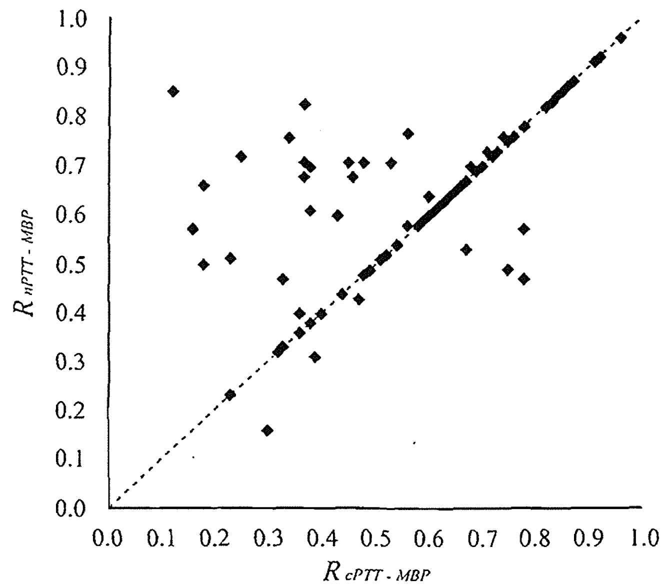

Fig. 8. Scatter diagram of $R_{cPTT\text{-}MBP}$ versus $R_{nPTT\text{-}MBP}$. $nPTT$ was selected from $cPTT$ or $SPTD$ on the basis of the PR value

of $cPTT$ increased the value of $R_{nPTT\text{-}MBP}$ for 25 subjects (placed above the dashed line), whereas this value decreased for 7 subjects (placed below the dashed line).

The accuracy of detecting the position at which the pulse wave begins to rise is important for estimating $cPTT$. Thus, noise contamination in the PPG signal, especially around these rising points, significantly reduces the accuracy of $cPTT$. On the other hand, the result shown in Fig. 8 suggests that $SPTD$ is more robust against short-term noise and artifacts in the PPG than $cPTT$, because $SPTD$ can be calculated on the basis of a global pattern of the PPG waveform.

4 Conclusions

In this study, we focused on the pulse transmission time (PTT), as a parameter containing information about blood pressure, to use in estimating the baroreflex characteristics without measuring a continuous blood pressure signal. To obtain a new parameter with a higher correlation with blood pressure in the Mayer wave-related band, rather than using the PTT only, we proposed the same-phase temporal difference (SPTD) based on ECG and PPG signals. Experiment on 94 healthy subjects using the combination of the SPTD and conventional PTT revealed that the correlation with blood pressure for 25 subjects could be improved by using the SPTD instead of the PTT, by applying the criterion of the ratio of their powers. However, the correlation decreased for 7 subjects.

In future work, we will calculate the baroreflex characteristics using the proposed parameter to confirm its validity. Furthermore, we will investigate why the correlation decreased in some cases. This investigation could lead to the development of an alternative method for extracting information with a higher correlation with blood pressure from the ECG and PPG signals.

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Estimation of Maximum Ventricular Elastance Under Assistance With a Rotary Blood Pump

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Abstract: The maximum ventricular elastance is a reliable index for assessing the cardiac function from changes in its pressure-volume relationship. The advantage of this index is that it can represent the contractility of either unassisted hearts or native hearts assisted with rotary blood pumps. However, there are situations in which changes in the ventricular load required for the conventional estimation method might be risky. For example, in a bridge-to-recovery the cardiac function should also be continuously

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observed after the implantation of a rotary blood pump. In this article, we present the results of the estimation of the maximum elastance with in vivo data using the parameter optimization method, which is a single-beat estimation method. The estimated values for the normal cardiac function (6.8 ± 0.6, 4.5 ± 0.9, 4.2 ± 1.8 mm Hg/mL) were significantly different from those for the low cardiac function (3.2 ± 1.5, 1.9 ± 1.0, 1.9 ± 1.2 mm Hg/mL) from the data of the three animals that were analyzed. Besides, the maximum elastance values were independent of the pump rotational speed. These results indicate that this index might be useful for the detection of the myocardial recovery. **Key Words:** Elastance—Estimation techniques—Heart contractility—Myocardial recovery—Ventricular assist device—Ventricular function.

Rotary blood pumps (RBPs) have been used as bridge-to-transplantation or as destination therapy for a relatively long period, and lately have also been used as bridge-to-recovery. In the latter case, the RBP unloads the native heart while the myocardial repair is carried out by pharmacological treatments or cell therapy. The RBP is then withdrawn when sufficient recovery of cardiac function is detected. Since the withdrawal and the reimplantation, in particular, of the RBP are invasive procedures, this recovery must be closely monitored before the weaning of the RBP (1).

Currently, in order to observe the recovery of the cardiac function, the pump must be temporarily stopped (1) so that the physiologist can analyze the myocardial response. Although the off-pump tests are accurate, stopping the pump is a risky procedure, during which there can be thrombi formation or insufficient blood perfusion in peripheral tissues. Thus, a safer technique for assessment of the cardiac function is necessary.

Many indices have been used to detect the function of the unassisted native heart, such as the ejection fraction (EF), the dP/dt_{max} and the maximum ventricular elastance (E_{max}). However, the RBP changes the hemodynamics of the cardiovascular system. Therefore, it is important to verify the validity of such indices during the assistance with RBPs.

The main objective of this study was to evaluate the validity of cardiac function indices during the assistance with RBPs. In particular, we choose the E_{max} to be analyzed as it represents the ventricular contractility independently of the load (2–4), which was expected to compensate for the changes brought by different assistance conditions. In order to avoid sudden and unsuitable changes of the ventricular load that are required for the conventional multiple-beats estimation method, we present an evaluation for the estimation of E_{max} using the parameter optimization method (POM), a single beat estimation method (5). As the gold standard, the E_{max} was also estimated using the conventional method during the off-pump condition.

MATERIALS AND METHODS

This study was performed in three healthy adult goats (female, 54, 52, and 58 kg) with the left ventricle assisted by the centrifugal pump NEDO PI-710 gyro pump (Baylor College of Medicine, Houston, TX, USA) in animals 1 and 2, and by the centrifugal pump EvaHeart (Sun Medical Technology Research Corporation, Nagano, Japan) in animal 3. The RBP outflow cannula was anastomosed to the descending aorta and the inflow cannula was inserted into the left ventricular apex. All animals received humane care in accordance with the guidelines determined by the Institutional Animal Care and Use Committee of Tohoku University. Ultrasonic flow meters (Transonic Systems, Inc., Ithaca, NY, USA) were placed on the pump outflow cannula and on the ascending aorta. A conductance catheter (Leycom, The Netherlands) was inserted into the left ventricle through the ascending aorta for continuous monitoring of the left ventricular pressure (LVP) and volume (LVV). Pump rotational speed and motor power supply were monitored through the controller. Heart failure conditions were mimicked by the injection of propranolol. The drug dose was determined according to the animal's condition at the time of the injection (respectively, 1.7, 0.25, and 2 mg). Each data set was recorded for 60 s with constant mean rotational speed at a sampling frequency of 1 kHz, starting at least 60 s after any changes in the rotational speed for the stabilization. For each animal, 12 data sets were recorded: six sets recorded at the control condition (hereinafter referred to as normal cardiac function [NCF]) and six sets recorded after the propranolol injection (hereinafter referred to as low cardiac function [LCF]); at each cardiac condition, one set was recorded with the pump stopped and the outflow cannula clamped, while the other five sets were recorded each one with a different mean pump rotational speed. The data with the pump stopped was recorded during manual clamp of the aorta, which changed the ventricular afterload. This data was used for the estimation of E_{max} using the multiple-beats method, approximating the E_{max} to the end-systolic elastance (E_{ES}), which corresponds to the slope of the end systolic pressure volume relationship (ESPVR) (2–4). The other data were used for the estimation of E_{max} using the POM, a method in which the E_{max} is estimated at each cardiac cycle independent of changes in the ventricular load. This method considers 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.

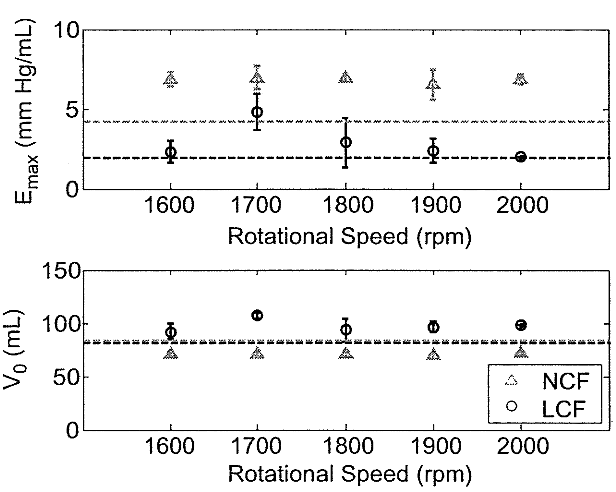

FIG. 1. (a) E_{max} and (b) V_0 comparing the conventional estimation method (dashed line) and POM (markers, mean ± standard deviation) for different pump rotational speeds using the data from animal 1. The black lines and markers represent the low cardiac function (LCF) and the gray lines and markers,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

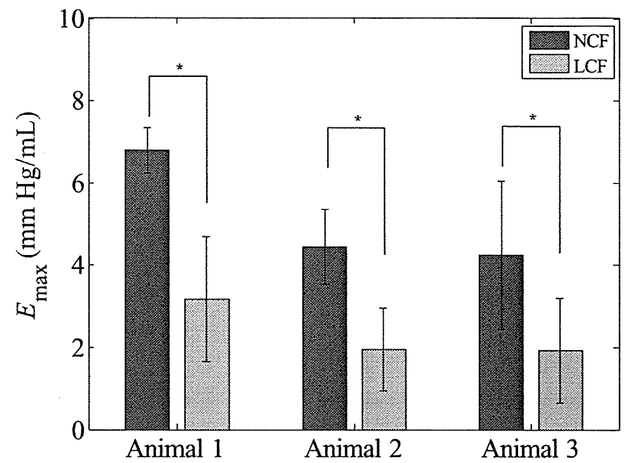

FIG. 2. E_{max} estimated for each cardiac function, independent of the pump rotational speed, represented by the mean and the standard deviation. $^*P < 0.01$ between NCF and LCF.

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