Assessing PEP and LVET from Heart Sounds: Algorithms and Evaluation

R. P. Paiva, P. Carvalho, X. Aubert, J. Muehlsteff, J. Henriques and M. Antunes

Abstract—**This paper addresses the estimation of systolic time intervals, namely the pre-ejection period (PEP) and the left ventricular ejection time (LVET), using heart sound. PEP is estimated with a Bayesian approach resorting to the signal's instantaneous amplitude and typical time intervals between atrio-ventricular valve closure and aortic valve opening. As for LVET, aortic valve closure is determined through the analysis of a high-frequency signature of S2. Additionally, LVET has also been estimated from a PPG signal at a peripheral site, for the sake of comparison over a subset of data. We evaluated our algorithms on a set of 658 heartbeats and achieved 10.32 msec average absolute PEP estimation error with 7.3 msec standard deviation and for LVET, 15.8 msec average estimation error with 13.6 msec standard deviation. Current results support our assumption that heart sounds can be applied to detect the onset of the aortic valve movement processes.**

I. Introduction

CARDIAC reserve parameters, such as contractility or cardiac output, provide crucial information regarding cardiovascular state. However, current measurement methods are usually invasive, expensive, require specific skills and, hence, are not performed customarily. For these reasons, recent research is being carried out aiming at non-invasive, cheaper and flexible techniques, able to accurately determine those parameters. Moreover, such techniques would be valuable, e.g., for patient home monitoring.

Several studies [1-3] have shown that cardiac systolic and diastolic time intervals are highly correlated to major and fundamental cardiac functions. Of major relevance in assessing the cardiac reserve and the left ventricular function are the pre-ejection period (PEP) and the left ventricular ejection time (LVET) [4-6].

By definition, PEP is the time interval between the start of ventricular depolarization and the moment of aortic valve opening. Nevertheless, for detection accuracy, PEP is defined in this paper as the interval between the R-peak of the electrocardiogram (ECG) and the opening of the aortic valve. As for LVET, it is defined as the time interval of left ventricular ejection, which occurs between the opening of the aortic valve and its subsequent closure. PEP is an index of the left ventricular function and reflects changes in myocardial contractility, left ventricular end-diastolic volume and aortic diastolic pressure. Another important application of PEP is in non-invasive beat-by-beat estimation of blood pressure [6]. The left ventricular ejection period (LVET) can also be related to contractility and to cardiac output. It is by itself a measure of cardiac function.

The current clinical gold standard method for assessing LVET and PEP is echocardiography. However, this is not feasible for patient home monitoring, where portable devices for non-invasive and low intrusive beat-by-beat measurements are needed. Hence, several measurement principles are being considered in the literature, ranging from oxygen saturation, radial pulse pressure [7] and impedance cardiography (ICG) [8]. ICG is indeed one of the reference methods for portable devices in measuring these parameters. However, as stated in [9], there is evidence that ICG does not enable the detection of the onset of the aortic valve opening and closing process. In fact, in [9], PEP values extracted from the ICG using a visual inspection method for B-point determination based upon the dZ/dt are delayed by 3-20ms relative to the onset of blood flow in the left ventricular trace, in comparison to the values extracted from echocardiographies.

Concerning LVET, several methods have been used prior to ultrasound deployment for detecting the ventricular ejection timing from pulse wave signals recorded at a peripheral site. This includes digital processing of the carotid pulse and ear dendrogram [e.g., 10]. A recent paper [11] proposes an elaborate method based on high-order derivative analysis of a photo-plethysmographic (PPG) signal recorded at the finger, which achieves remarkable accuracy for LVET estimations far away from the heart.

In this article, the goal is to describe algorithms for accurately extracting the systolic time intervals (PEP and LVET) using heart sound (HS) and ECG. The underlying hypothesis is that the first and the second heart sounds encode the movements of the aortic valve and that these components exhibit noticeable and specific signatures that enable their identification using this signal. In an attempt to validate this hypothesis, we conducted a feasibility study, which confirmed that the opening and closing events of the aortic valve could be extracted from the first and the second heart sounds, respectively. This is reported in [12].

Our PEP estimation algorithm was evaluated on a set of 658 heartbeats, achieving 10.32 msec average absolute error, with 7.3 msec standard deviation. As for LVET estimation from heart sound, we were only able to use 333 beats, due to annotation difficulties of aortic valve closure, from which

This work was supported in part by the EU FP7 project HeartCycle (FP7 – 216695).

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15.8 msec average estimation error and 13.6 msec standard deviation resulted. LVET estimations from a PPG finger clip achieve comparable results, however, over a subset of test data involving only four subjects and 112 heartbeats.

The paper is organized as follows. In sections II and III, the algorithms for PEP and LVET estimation are described. In Section IV, experimental results are presented and discussed. Finally, conclusions from this study are drawn in Section V.

II. PEP Estimation

We follow a Bayesian approach for PEP estimation, resorting to the instantaneous amplitude (IA) of the heart sound waveform as the main feature. The motivation to this approach comes from the fact that the closure of atrioventricular (AV) valves is usually easy to detect, as it corresponds to a strong peak in the IA. We then analyze the IA curve and estimate the PEP duration based on the typical delay between AV closure and aortic valve opening found in the literature [4]. The PEP value of the previous heartbeat is also included in the model to somewhat constrain the range of possibilities since we assume that, at rest, abrupt variations are not likely to occur.

Before PEP estimation, we apply an in-house algorithm developed for R-peaks detection from the ECG [13].

The algorithm for PEP estimation then starts by determining the signal's instantaneous amplitude, $a(t)$, via the analytic signal, $s_a(t)$, as in (1). There, HT denotes the Hilbert Transform:

$$\begin{aligned} a(t) &= |s_a(t)| \\ s_a(t) &= s(t) + jHT(s(t)) \end{aligned} \tag{1}$$

Next, we estimate the AV closure time interval with reference to the corresponding previously determined R-peak in the current heartbeat. A Bayesian model is defined, where the IA curve and the previous AV interval are employed, according to (2):

$$p(AV_k \mid IA_k, AV_{k-1}) \approx \frac{p(AV_k \mid IA_k) \cdot p(AV_k \mid AV_{k-1})}{p(AV_k)} \tag{2}$$

In this equation, k stands for the heartbeat number. In (2), $p(AV_k \mid AV_{k-1})$ is modeled as a Gaussian distribution centered in the previous AV interval and with a standard deviation of 20 msec. Also in (2), we define $p(AV_k \mid IA_k)$ = normalized(IA_k), given the assumption that higher amplitude values are more likely to correspond to AV closure. Moreover, we assume uniformity for the AV_k distribution.

We then estimate the AV interval as the maximum of $p(AV_k \mid AV_{k-1}, IA_k)$. Here, it is important to notice that, in order to improve the robustness of the model to estimation errors in previous heartbeats, we keep track of the AV distribution in the previous beat and test all possible AV time intervals. These are weighted by the corresponding individual probabilities.

After AV closure interval estimation, PEP duration is inferred. Again, we follow a Bayesian strategy, resorting to the IA curve, the estimated AV interval and the previous PEP duration, as follows (3):

$$\begin{aligned} & p(PEP_k \mid AV_k, IA_k, PEP_{k-1}) \approx \\ & \approx \frac{p(PEP_k \mid AV_k) \cdot p(PEP_k \mid IA_k) \cdot p(PEP_k \mid PEP_{k-1})}{p(PEP_k)} \end{aligned} \tag{3}$$

In (3), $p(PEP_k \mid PEP_{k-1})$ is also modeled as a Gaussian distribution centered in the previous PEP duration with a standard deviation of 30 msec. In the same equation, we define $p(PEP_k \mid IA_k)$ = 1 - normalized(IA_k), since aortic opening seems to correspond to valleys in the IA curve, as we have observed in the conducted experiments. Regarding the $p(PEP_k \mid AV_k)$ distribution, this one is modeled as a Gaussian centered in AV_k + 30 msec, once again with a standard deviation of 30 msec. This was motivated by results found in the literature, which indicate that the aortic valve opens typically 30 msec after the closure of AV valves [4]. Finally, as before, we assume uniformity for the PEP_k distribution.

Once again, we keep track of the PEP distribution in the previous beat and test all possible PEP durations in order to improve model reliability.

An example of the obtained AV closure and PEP probability distributions for one heartbeat are shown in Fig.1.

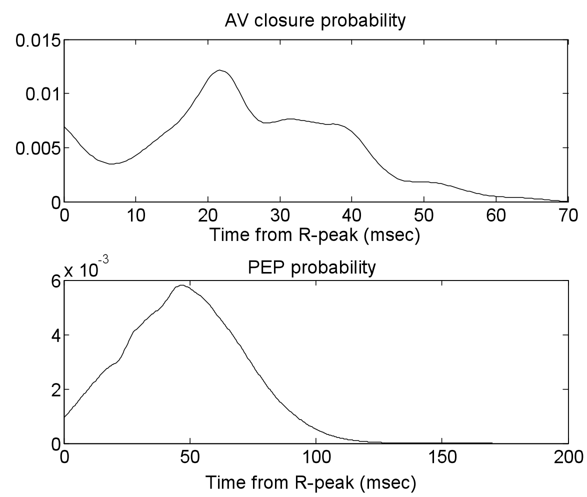

Fig. 1. AV closure and PEP probability distributions.

III. LVET Estimation

A. Heart Sound

The proposed method for LVET estimation using HS is based on the observation that the frequency of valve vibration depends on the pressure difference across the valves and is an adaptation of our method reported in 2006 [2], which was originally proposed for heart sound segmentation.

From the knowledge of cardiac functionality and genesis of S1 and S2 sounds, it is known that aortic valves close with relatively large pressure difference across the valve. This high-pressure difference justifies the high frequency mingling in S2 sounds. Thus, usually S2 sounds contain higher frequency content compared to S1 sounds (excluding some rare exceptions, such as for some prosthetic valve implants). Nevertheless, this characteristic may be used as a marker, much in the same way as the QRS-complex in ECG, to identify the second heart sound.

Briefly, in order to find the presence of high frequency information in at least one type of heart sound, detail coefficients of the Fast Wavelet Transform (FWT) are considered. To extract the high frequency envelopes in sound segments, the Shannon energy operator is applied to the detail coefficients. In order to detect the heart cycles, an adaptive threshold is defined for this Shannon energy envelope. Further details on the algorithm for aortic valve closure detection can be found in [2].

This algorithm was originally developed in the context of heart sound segmentation, for application in the detection of prosthetic heart valve dysfunctions. As this problem does not entail the same temporal constraints, a more conservative approach was followed, as to the accurate detection of segment starts and endings. Hence, here we improved the algorithm by redefining the start of each S2 sound as the point where the signal's energy reaches 10% of the maximum energy in the segment.

Finally, LVET is simply obtained by calculating the difference from aortic closing and opening times (this one obtained from the PEP estimation algorithm).

Fig. 2 illustrates the analysis for one typical S2 sound segment. There, the solid function plot is the signal energy, while the dashed function plot represents the low-pass filtered Shannon energy. The three vertical lines are, from left to right: initial definition of the S2 sound start, based on the Shannon energy; corrected S2 sound start based on the signal energy; and the end of the S2 sound.

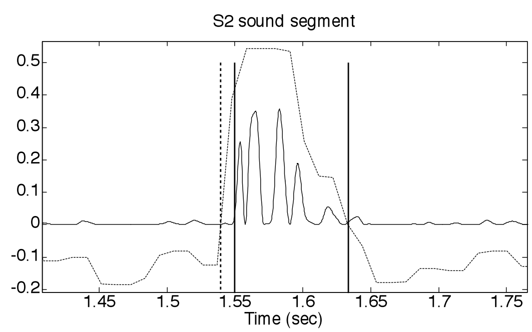

Fig. 2. S2 sound segmentation and start adjustment.

B. PPG

Similar to [11], the PPG pulse-waveform is analyzed based on its successive derivatives up to the fourth order, computed from a polynomial-fitting of the original signal. The systolic ejection onset is determined at the time achieving the maximum of the third derivative along the up-rising wave. The end of systolic ejection is computed from several features including the slope, curvature and third derivative. The LVET estimation is obtained with a rule-based decision logic, taking account of the morphology of the falling part of the pulse (e.g. presence or absence of a dichrotic notch).

IV. Experimental Results

A. Experimental Setup

We carried out a small data collection study involving 17 students at the Centro Hospitalar de Coimbra to simultaneously collect heart sounds and echocardiographies (Echo). The data acquisition process was conducted in 3 stages. In the last one, the PPG signal was also collected, from a population of 4 volunteers. A synchronous ECG with each of the above signals was also acquired and served as a reference signal for co-registration.

All subjects had no known congenital or other heart disease. The biometric characteristics of the population were:

- Age: 22.53 ± 3.81 years
- BMI: 23.27 ± 2.15 Kg/m2
- Heart rate: 72.94 ± 9.87 bpm
- 14 males and 3 females

The measurement protocol was performed by an authorized medical specialist and consisted of several acquisitions of echocardiography in different modes (Doppler and M-mode) and heart sound collection sites (apex and left sternum border). Details on the protocol are described in [12]. The PPG signals have been acquired from a finger clip sensor and a CMS Monitor from HP, at a sampling rate of 125 Hz.

After data acquisition, the annotations of the opening and closing instants of the aortic valve using the echocardiographies were performed under the supervision of an experienced clinical expert in echocardiography. The detected aortic opening and closing times, with the associated PEP and LVET values, were used as ground truth for algorithm evaluation.

B. Evaluation and Analysis

Our approach was evaluated in a set of audio clips from the echocardiography-HS collection data, with manually annotated aortic valve openings (based on synchronized echo-cardiography images).

Table I summarizes the achieved results. Regarding PEP estimation, 658 annotated beats were employed, from which 10.32 msec absolute average error, with 7.3 msec standard deviation resulted, i.e., 21.71% ± 15.36%, relative to the average annotated PEP values (47.55 msec). Moreover, 0.47 correlation (ρ) between annotated and estimated PEP values was obtained.

TABLE I
Summary of Results

Parameter	Annotated Range (msec) (average ± std)	Estimation Error (msec) (average ± std)	ρ
PEP	47.55 ± 12.72	10.32 ± 7.3	0.47
LVET HS	268.46 ± 24.26	15.8 ± 13.6	0.77
LVET PPG*	255.85 ± 18.24	11.5 ± 14.2	0.77

*Only applied over a subset of 112 beats where PPG was available.

Fig. 3 shows the PEP estimation difference dispersion as a function of the beat-by-beat values from the echocardiography and heart sound. The horizontal dashed lines denote standard deviation boundaries. As can be observed, the average error is close to zero (precisely, -0.65 msec), with a 12.63 msec standard deviation.

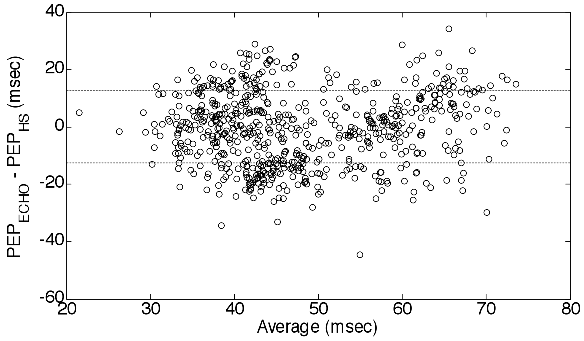

Fig. 3. PEP estimation difference dispersion as a function of the beat-by-beat ($PEP_{ECHO} + PEP_{HS}$)/2.

In order to assess the sensitivity of the algorithm to parameter variations, the mean difference between AV closure and aortic valve opening was varied up to ±15 msec from the nominal value. Also, the standard deviations (std) of all Gaussians were varied in the same range. As for std variations, these had nearly null impact in the results: the maximum observed average error was 11 msec. Regarding variations of the mean, these had a strong impact on the results as expected: a 45-msec mean average value led to 17.42 msec error. Thus, our results seem to confirm Tavel's indication that the aortic valve opens typically 30 msec after the closure of AV valves. However, it should be pointed out that this could change with cardiac pathology or load variations. Hence, we have employed a broad standard deviation in the Gaussian distribution to accommodate larger variations from the average.

There are several possible causes for the errors obtained. First of all, we used only one feature (IA), which, although being relevant, is insufficient to capture all the dynamics involved in the aortic valve opening process. Namely, frequency variations also occur, and so instantaneous frequency (IF) in combination with IA is likely to improve the results. Also, the signal-to-noise ratio (SNR) in several audio waves was very poor, the reason why we disposed of some samples. Anyway, SNR was still not ideal in several of the employed sound samples. Moreover, some errors also draw from synchronization difficulties (particularly due to lack of Echo-ECG resolution).

In any case, the PEP results are quite promising. In fact, they are very close to the results reported in our feasibility study [12]. However, it can be seen that there is still room for algorithmic improvements.

Fig. 4 illustrates typical results of PEP estimation for one heart beat. The vertical lines denote (from left to right): R-peak, AV closure, estimated and annotated aortic valve opening. As can be seen, the closure of AV valves is accurately detected (a strong peak in the IA curve) and the estimated aortic valve opening is close to the one annotated via the echocardiography (around 8 msec).

As for LVET estimation from HS, we have only used 333 beats, due to annotation difficulties of aortic valve closure. We achieved 15.8 msec average absolute error, with 13.6 msec standard deviation, i.e., 5.88% ± 5.07%. Correlation between annotated and estimated LVET values was 0.77.

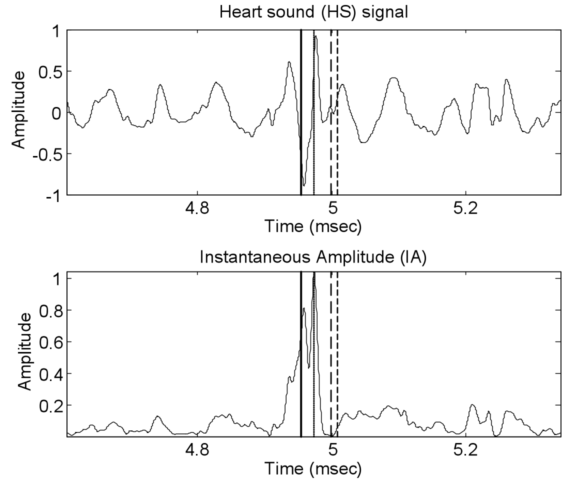

Fig. 4. Illustration of PEP estimation.

The LVET estimation difference dispersion as a function of the beat-by-beat values from the echocardiography and heart sound is shown in Fig. 5. The horizontal dashed lines represent standard deviation boundaries. Unlike PEP estimation, for LVET we obtained an average error of 10.21 msec (with 18.18 msec standard deviation), as can be observed in Fig. 5. In fact, the annotated LVET is characterized by a typical delay of around 12 msec, as discussed below.

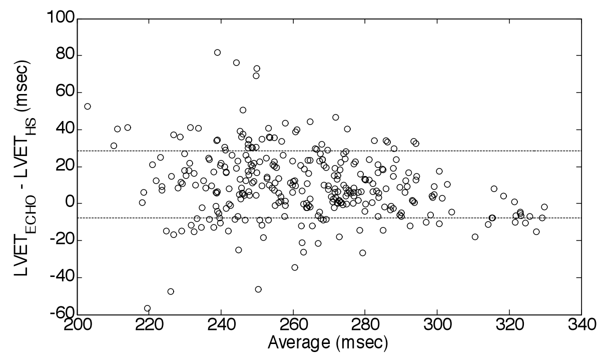

Fig. 5. LVET estimation difference dispersion as a function of the beat-by-beat ($LVET_{ECHO} + LVET_{HS}$)/2.

Again, the errors obtained stem from several sources. Besides the mentioned difficulties with SNR and signal synchronization, the annotated LVET is persistently delayed compared to the start of sound S2. In our feasibility study [12], it was observed that on average the onset of the aortic valve closing movement was detected 12.1 msec earlier compared to echocardiography. This can be attributed to the fact that HS enables the detection of the onset of the aortic valve closing process, while echocardiography enables its detection near the closing click induced by the valve cusps, i.e. at the end of the dynamic process. Also, LVET estimation error suffers from propagation of PEP detection errors. However, this effect is not as high as could be expected. In fact, if LVET is calculated based on the annotated PEP values, the error decreases only slightly to 15.12±12.85 msec.

Fig. 6 illustrates typical results for aortic valve closure detection via the extracted high frequency signatures in S2 sounds. The upper plot is the ECG. Down, the continuous line represents sound energy and the dashed line denotes Shannon energy. The vertical dashed lines represent segment starts and the continuous ones stand for endings. S2 segment starts correspond to aortic valve closures.

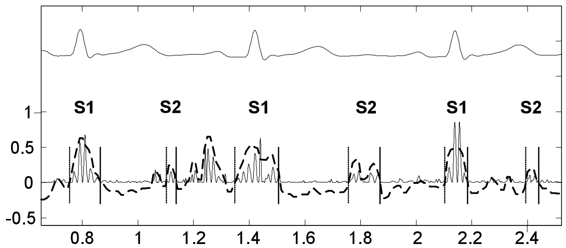

Fig. 6. Illustration of aortic valve closure detection.

The PPG-derived LVET estimations could only be obtained over a subset of data from a signal sampled at 125 Hz. Nevertheless, this algorithm provides fairly robust estimations and is able to track the beat-to-beat fluctuations of the LVET intervals, thus confirming the validity of using higher order derivatives obtained from the smoothed PPG signal. Current results show that the PPG-derived LVET values are comparable to those provided by the heart-sound analysis.

V. Conclusions

In this work, we studied the possibility of using heart sounds to measure the main systolic heart time intervals, i.e. the pre-ejection period and the left ventricle ejection time. The basis hypothesis was that heart sounds encode markers that enable the detection of the opening and closing instants of the aortic valve. For an objective evaluation, we performed an echocardiography-heart sound study over 17 healthy subjects. We employed the instantaneous amplitude of the heart sound as the main feature for PEP estimation. As for LVET, we compared results from heart sound, using a high-frequency signature, with the ones obtained from the PPG signal at finger and observed similar values. The achieved results support our hypothesis that heart sound can be applied to detect the onset of aortic valve movement processes, though they should be regarded as preliminary due to the limited amount of data and the early stage of evaluation.

As future work, besides improving our current algorithms (namely, employing also the instantaneous frequency) we will conduct a comparative study with other competing approaches, namely the ICG-based methodologies. These tend to exhibit biases in the determination of the considered systolic time intervals, leading to inaccuracies in cardiac function assessment, which may be reduced using heart sounds.

In addition, we plan to extend this study with new measurements from a population with coronary heart diseases, besides the healthy population we have considered so far.

Acknowledgment

The authors want to express their gratitude to the 17 students from the Centro Hospitalar de Coimbra (CHC) who volunteered for this study. The authors would also like to recognize and to express their appreciation to the Centro Hospitalar de Coimbra for supporting the study. The effort of Dr. Leitão Marques and Dr. Isabel Quintal from CHC in facilitating the arrangements for the data acquisition part of the study is also recognized.

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