

Automatic Extraction of Physiological Features from Vibro-Acoustic Heart Signals: Correlation with Echo-Doppler

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Abstract

The mechanical processes within the cardiovascular system produce low-frequency vibrations and sounds. These vibro-acoustic signals carry valuable physiological information that can be potentially used for cardiac monitoring. In this work, heart sounds, apical pulse, and arterial pulse signals were simultaneously acquired, along with electrocardiogram and echo-Doppler audio signals. Processing algorithms were developed to extract temporal and morphological feature from the signals. Spectral analysis was used to reconstruct the Doppler sonograms and estimate reference values. A good agreement was observed between systolic and diastolic time intervals estimated by both methods. Strong beat-to-beat correlations were shown both in rest and during pharmacological stress test. The results demonstrate the technological and medical feasibility of using automatic analysis of vibro-acoustic heart signals for continuous non-invasive monitoring of cardiac functionality.

1. Introduction

The mechanical functionality of the cardiovascular system is governed by a complex interplay between pressure gradients, determined by the contraction force of the myocardial cells, the dynamics of blood flow and the compliance of cardiac chambers and blood vessels. These mechanical processes produce vibrations and acoustic signals that can be recorded over the chest wall. Vibro-acoustic heart signals, including heart sounds (phonocardiogram), apical pulse (apexcardiogram) and arterial pulse (e.g. carotid pulse) carry valuable clinical information, but their use has been mostly limited to qualitative assessment by manual methods [1].

The two major audible sounds in a normal cardiac cycle are the first and second heart sounds, S1 and S2. S1 occurs at the onset of the ventricular contraction during the closure of the mitral and tricuspid valves. S2 is heard at the end of the ventricular systole, during the closure of the aortic and pulmonary valves. The apexcardiogram

signal reflects the low-frequency vibrations transmitted by the heart and great vessels to the chest wall at the point of maximal impulse. The carotid pulse signal reflects small volume changes of the artery with each heartbeat. The morphology of these signals and their temporal relationships are illustrated in Figure 1.

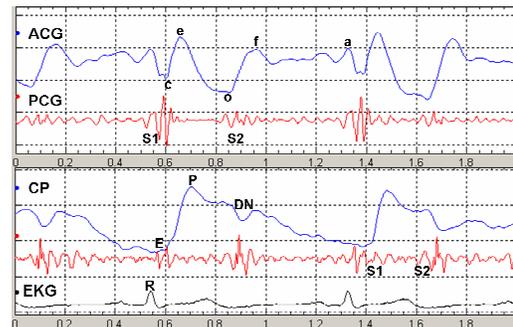


Figure 1: Vibro-acoustic heart signals: Simultaneous traces of apexcardiogram (ACG), carotid pulse (CP), phonocardiogram (PCG) and electrocardiogram (EKG).

The clinical application of vibro-acoustic heart signals have been extensively studied 30-40 years ago [2][3]. Analysis techniques, including time-domain, frequency-domain and modeling methods, have been developed for heart sounds [4]. However, with the introduction of modern imaging techniques, much of the recent research on physiological heart signal analysis has focused on electrical and ultrasonic signals, whereas mechanical and acoustic signals were relatively overlooked. The motivation for this work is to revisit automatic analysis of mechanical heart signal using modern signal processing algorithms, and to demonstrate the feasibility of using such algorithms to extract features that reliably represent the underlying physiological processes.

2. Methods

2.1. Data acquisition

A digital data acquisition system, able to simultaneously acquire multiple sensory modalities, was constructed in our lab. The system is consisted of piezoelectric contact transducers (PPG Sensor Model 3,

OHK Medical Devices), an EKG sensor (EKG-BTA, Vernier Software & Technology), a preamplifier with high input impedance and a linear frequency range of 1Hz – 4KHz, a 16-bit analog-to-digital converter (PMD-1608FS, Measurement Computing Corp.), and a recording software written in Matlab 6.5 (The Mathworks Inc.), running on a mobile personal computer.

The system was used to simultaneously record carotid pulse (*CP*), apexcardiogram (*ACG*), phonocardiogram (*PCG*), 3-lead electrocardiogram (*EKG*) and echo-Doppler audio signals from 5 healthy volunteers. In addition, data was recorded from a cardiac patient during Dobutamine stress echo test. Subjects were recorded in the left-lateral supine position. The vibro-acoustic transducers were firmly attached using elastic straps. Doppler echocardiography was acquired with an Acuson Sequoia C256 (Siemens Medical Solutions). The echo transducer was placed at the apical window, and the acquired data was sampled at a rate of 8000 Hz.

2.2. Doppler-audio processing

The pattern of mitral and aortic blood flow was reconstructed from the Doppler audio signals by applying short-time Fourier transform (with a Hamming window of 256 samples). The scatter noise level was estimated from the spectrum, and velocities with amplitude below the noise level were filtered-out. The instantaneous flow through the mitral and aortic valves was estimated by integrating the positive and negative velocities, respectively, at each time stamp, followed by smoothing with a moving-average filter (Figure 2). The resulting profiles were shifted backwards in time, to compensate for the constant delay imposed by the processing time of the echocardiograph. The temporal location of the aortic ejection wave and the mitral filling waves were manually annotated in the instantaneous flow profiles. These annotations were used to estimate the reference systolic and diastolic time intervals and the ejection and filling amplitudes.

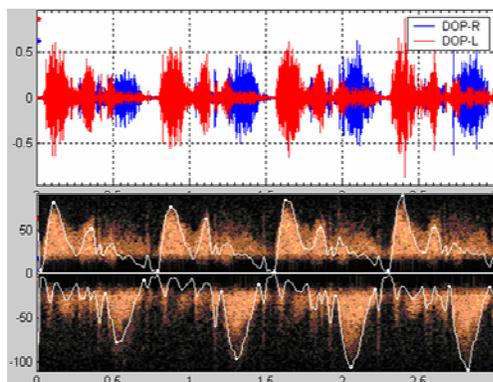


Figure 2: Doppler-audio waveforms (a) are processed by short-time Fourier transform to get an instantaneous flow profile that is manually annotated (b).

2.3. Signal segmentation

The recorded signals were partitioned into cardiac cycles by identifying the peak of the QRS complex in the EKG signal. For each cycle, an automatic segmentation algorithm was applied to identify the distinct components of the vibro-acoustic signals, and determine the derived timing of cardiac events. The segmentation algorithm first identifies S1 and S2 in the PCG signals, then uses the temporal location of the heart sounds to localize the heuristic search for distinct extrema points in the pulse signals (*ACG*, *CP*). The algorithm steps are as follows:

1. Identify S1 and S2 in each PCG signal:
 - 1.1. Compute the amplitude envelope of the PCG signal
 - 1.2. Locate the dominant peaks of S1 and S2
 - 1.3. Adjust the locations of S1 and S2 by reducing their temporal variability
2. Identify the points of ejection (*E*), peak (*P*) and diastolic notch (*DN*) in the *CP* signal:
 - 2.1. Locate *E* as a local minima in the proximity of S1 and the maxima of the signal's second derivative
 - 2.2. Locate *P* as a local maxima between S1 and S2
 - 2.3. Locate *DN* as a local minima in the proximity of S2
3. Identify the points of contraction (*c*), ejection (*e*), relaxation (*o*), filling (*f*) and atrial contraction (*a*) in the *ACG* signal:
 - 3.1. Locate *e* as a local maxima in the proximity of S1
 - 3.2. Locate *c* as a local minima preceding point *e*
 - 3.3. Locate *o* as a local minima in the proximity of S2
 - 3.4. Locate *f* as a local maxima following point *o*
 - 3.5. Locate *a* as a local maxima preceding EKG R-wave

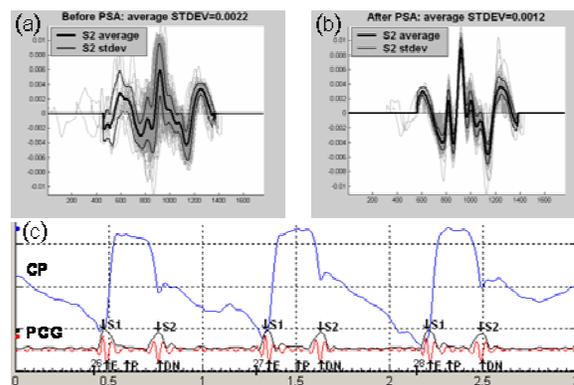


Figure 3: Segmentation of heart signals: The variability of S2 signals before (a) and after (b) applying PSA. CP segmentation using the temporal locations of S1 and S2 (c).

The amplitude envelope of the PCG signal is obtained by applying a low-pass filter on the instantaneous amplitude of the analytic signal $z(t)=s(t)+iH[s(t)]$, where $s(t)$ is the PCG signal and H is the Hilbert transform. The initial temporal location of S1 or S2 is determined by the dominant peaks in the PCG envelope. To adjust the location of the sound signals, a phase-shift averaging technique is applied: (1) The signals of multiple cycles

are aligned according to the initial peak. (2) The ensemble average of the multiple cycles is calculated. (3) The phase shift that maximized the correlation between each cycle and the average is computed and the cycles are shifted accordingly. (4) A new average signal of the shifted signals is calculated, and its strongest component is identified. (5) The phase shift of each cycle with regard to the strongest component of the average is calculated, and the final location of S1/S2 in each cycle is determined. Figure 3 shows an example for improving the variability of S2 by PSA, and using the PCG for CP segmentation.

2.4. Feature extraction

Temporal and morphological features, representing the underlying physiological processes, were extracted from each cycle of the segmented vibro-acoustic signals. These features included systolic time intervals of pre-ejection period (*PEP*) and ejection time (*ET*), diastolic time intervals of isovolumetric relaxation time (*IVRT*) and filling time (*FT*), duration between S1 and S2, and amplitude and average slope of the rapid ejection. For each of the features, the corresponding Doppler profile was annotated, and the reference estimations of the feature values were derived.

3. Results

The digital data acquisition system was able to simultaneously record electrical, acoustic, infra-sound and Doppler signals continuously and reliably. The Doppler sonograms, reconstructed from the Doppler audio signals, matched the Doppler traces recorded on video. Thus, the technological feasibility of the system was ascertained. Several representing examples of the relationship between the vibro-acoustic signals and the echo-Doppler reference are given in the next sections.

3.1. Timing of cardiac events

The temporal relationship between the carotid pulse signal and the pattern of aortic blood flow, obtained by Continuous-Wave Doppler is illustrated in Figure 4a. The rapid upstroke in the CP signal, indicating the beginning of rapid ejection, coincides with the beginning of the Doppler ejection wave. The aortic valve closure (S2) and the end of the Doppler ejection wave. Figure 4b also demonstrates the temporal relationship between the apexcardiogram signal and the pattern of lateral ventricular wall movement, obtained by Tissue-Doppler Imaging. The low-frequency content of the apex pulse during diastole matches the two negative Doppler waves of early ventricular relaxation and late atrial contraction. Thus, information fusion from both pulse signals, as well as from sound signals, allows to locate events of the cardiac cycle and to measure the duration of the cardiac phases with high accuracy.

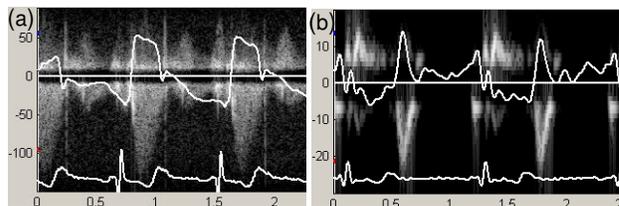


Figure 4: The relationships between the CP signal and CW-Doppler of aortic blood flow (a) and between the ACG signal and TDI of the lateral wall (b).

The average values of the systolic and diastolic time intervals, estimated from the pulse signals and the reference CW-Doppler profile of a healthy subject are given in Table 1. A good agreement in both average and standard deviation values was observed between the time intervals measured by both methods. The beat-to-beat filling time, derived automatically from the apexcardiogram signal, showed a strong statistical correlation ($r=0.92$, $p<1e-40$) with the Doppler-derived filling time. The average difference between the instantaneous filling times measured by the two methods was 0.77 ± 11.6 ms (Figure 5). Statistically-significant correlations were also observed for the ejection time and ejection amplitude, derived automatically from the carotid pulse signal. However, as the physiological variations of these measures are very low in rest conditions, the inherent measurement errors of both methods mask the small beat-to-beat physiological changes, thus weakening the correlation between the instantaneous measurements. Dynamic modification of cardiac functionality is therefore required to elucidate the correlation of systolic features. Such conditions can be achieved during dynamic exercising or pharmacological stress test.

Table 1: Systolic and diastolic time intervals (ms) derived from apexcardiogram (ACG), carotid pulse (CP) and CW-Doppler of 109 heart cycles recorded from a healthy subject.

Time interval	ACG	CP	Doppler
PEP	59.4 ± 1.8	62.4 ± 5.7	66.4 ± 5.9
ET	268.2 ± 4.1	262.2 ± 7.3	262.1 ± 7.1
IVRT	88.9 ± 10.1	-	90.1 ± 9.3
FT	313.1 ± 29.7	-	312.3 ± 30.6

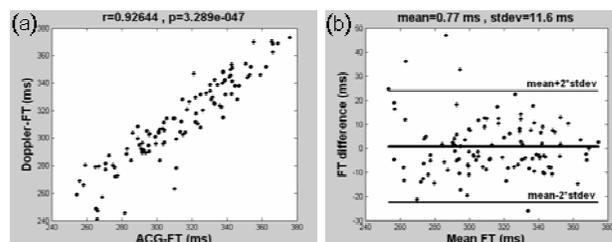


Figure 5: Correlation (a) and statistical agreement (b) between filling time derived from apexcardiogram and CW-Doppler of 109 heart cycles recorded from a healthy subject.

3.2. Stress test effects

Preliminary recordings were taken during Dobutamine stress echo examination of a cardiac patient. Dobutamine augments both heart rate and contractility, and raises the systolic blood pressure. The dynamics of heart rate and arterial pressure during the test are shown in Figure 6a and 6b. Since the patient was taking beta-blockers, Atropine was administered to reach the target heart rate. At seven points during the test, CW Doppler of the aortic valve blood flow was recorded. Figure 6c shows the average values of ejection time, derived from the Doppler reference, along with estimations of ejection time calculated from the delay between the major components of S1 and S2. The two estimations of ejection time showed strong correlation with each other ($r=0.95$, $p<1e-10$) and with the heart cycle period ($r=0.92-0.96$, $p<1e-9$). The relative magnitude of the ejection was estimated from the maximal amplitude of S1 and the average slope of the carotid-pulse upstroke (Figure 6d). The two estimations, taken from two different type of signals, were strongly correlated ($r=0.83$, $p<1e-6$). Significant correlations were found between the amplitude of S1 and the mean arterial pressure ($r=0.77$, $p<1e-4$) and between the CP average slope and the diastolic arterial pressure ($r=0.75$, $p<1e-4$).

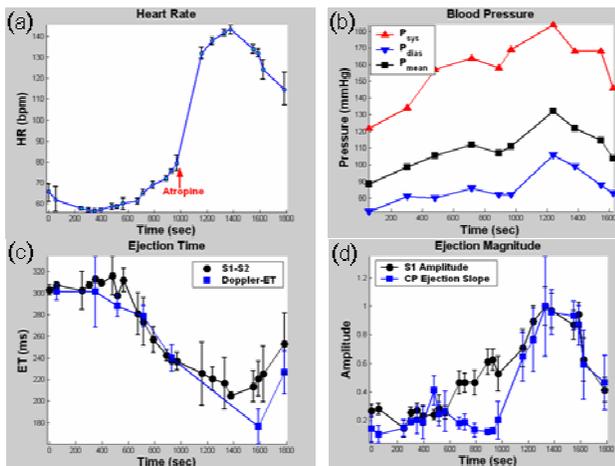


Figure 6: Heart rate (a), arterial pressure (b), estimated ejection time (c) and ejection magnitude (d) recorded during Dobutamine stress echocardiography test.

4. Discussion and conclusions

The results demonstrate the technological feasibility and the medical potential of extracting quantitative physiological information from vibro-acoustic heart signals. A good agreement with the reference echo-Doppler estimations was shown. While some of the variability observed in the extracted features represents genuine physiological changes, some of it is a result of measurement errors. When the physiological variation is large relative to the measurement errors, the strong

correlation between the estimations of the two methods emerges. An example is the diastolic filling time, which continuously varies with the instantaneous heart rate, even in rest conditions. However, for features with small physiological variability such as systolic ejection time, the measurement errors are masking the small beat-to-beat changes, and dynamic stress conditions are required to uncover the relationship between the estimations of the two methods.

The road towards fully-automated analysis of vibro-acoustic heart signals is scattered with pitfalls and challenges. The main technical difficulties include handling noise and artifacts that contaminate the acquired data, improving the algorithmic robustness with regards to the precise placement and attachment of the transducers, and ensuring the accuracy and reliability of the 'gold-standard' reference measurements. Future work will include further data collection for establishing the preliminary results, as well as identification and extraction of more complex features, which can be used to estimate key indices of cardiac mechanics, such as contractility, relaxation and compliance.

In conclusion, the vibro-acoustic heart signals, being a direct manifestation of mechanical processes within the cardiovascular system, can be used to estimate both the temporal and morphological properties of these processes. We believe that modern signal processing algorithms, along with recent technological advancements in medical sensors, computing devices and wireless communication, enable us to revisit these fundamental heart signals. Automatic algorithms for real-time analysis of vibro-acoustic heart signals can set the grounds for the challenging task of non-invasive continuous monitoring of the cardiovascular mechanical functionality.

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