Impact of the auscultation area on heart sound waveforms for the estimation of Cardiac Time Intervals

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Impact of the auscultation area on heart sound waveforms for the estimation of Cardiac Time Intervals

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Abstract*— Although clinicians still regard traditional auscultation as a key bedside assessment tool, interest in its digital counterpart has recently grown. Digitalizing heart sounds enables the extraction without the need of echocardiography of novel clinical indicators like Cardiac Time Intervals (CTIs), temporal parameters reflecting the hemodynamic behavior of the heart. The estimate of the CTIs grounds on the identification of the time of closure of heart valves, which the human ear cannot perceive - but algorithms can. Traditional auscultation areas are still considered a gold standard, but their suitability for CTI estimation was never thoroughly explored. In this study, we analyze the spatial variability of heart sound waveforms over the chest with high spatial resolution. We estimate the time of closure of cardiac valves and define a goal-oriented objective function to automatically identify the best auscultation area for each valve. We compare the selection against visual inspection (manually selecting the signals where discriminating between the left- and right-heart contributions was possible) and against the traditional auscultation areas. We found that the spatial similarity pattern of the sound waveforms recorded using the multi-source sensor is consistent with expectations. The variation in morphology produces a variation in the estimates, whose standard deviation ranges from 7 to 13 milliseconds, potentially causing issues in clinical interpretation. This proves that the selection of the recording point impacts the estimate and must be carefully defined. The estimates resulting from the automatic channel selection present higher correlation coefficients against visual selection (R = 0,73 to 0,96) than against traditional auscultation (R = 0,49 to 0,92). We can conclude that the definition of the best auscultation areas should be considered a goal-dependent task and that the spatial variability of heart sounds plays a role in obtaining robust estimates of clinical biomarkers.

***Clinical Relevance*— The reported spatial variation of the heart sound waveforms makes the definition of the best auscultation areas a goal-dependent task that should be adapted to the clinical task.**

I. Introduction

Digital auscultation, i.e., the digital recording of heart sounds, is gaining interest in scientific and medical communities as a noninvasive, portable, low-cost alternative for cardiovascular screening [1], [2]. It can be regarded to as an evolution of traditional auscultation, performed with an analog stethoscope, which is still considered an invaluable bedside clinical assessment tool despite its intrinsic limits. Digital auscultation enables the extraction of quantitative parameters from heart sounds and the collection of information that the human ear cannot perceive [3]. In this sense, it strongly enhances the potentiality of heart sounds as a monitoring and diagnostic tool. For example, digital phonocardiography (PCG) enables a temporal analysis of the cardiac cycle and the accurate estimation of the Cardiac Time Intervals (CTIs) [4] [5] [6],without the need of echocardiography, which reflect the hemodynamic behavior of the heart and proved a valuable tool to assess its electromechanical coupling noninvasively [7] [8] [9] [10] [11].

While listening to the patient's heart using a stethoscope, clinicians typically focus on murmurs and clicks, respectively indicating valvular regurgitations, valvular prolapse or stenosis. The presence of these acoustic features can be detected by a trained ear. Decades of clinical expertise in traditional auscultation led to the definition of standardized auscultation areas [12]: for example, the "tricuspid area" is located on the 4th left intercostal space along the sternal border because in this area the murmur indicating tricuspid regurgitation has the highest intensity.

Digital auscultation may have a variety of different goals, which only partially overlap with traditional, analog auscultation. If automatic murmur detection is definitely of interest, digital PCG opens to completely novel possibilities such as CTI estimation. When it comes to novel goals, the traditional methods do not necessarily yield the best results. In this case, our focus is on the auscultation area. Most previous works describe PCG recordings from the four traditional auscultation areas. Also, most publicly available datasets provide signals recorded over the traditional auscultation areas. Nevertheless, to our best knowledge, no previous work analyzed the impact of this choice on the parameters of interest when analyzing time-related features of the cardiac cycle.

In this study, we explore the impact of the auscultation area on the heart sound waveform and on the identification of the time of closure of four cardiac valves, aimed at estimating the CTIs. A more in-depth knowledge of the differences in heart sounds recorded from different points of the thorax can guide

* This research was funded by Compagnia di Sanpaolo through a Proof-of-Concept grant (PoC Instrument 2022-2024, Transition line, first cutoff).

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the clinician in the best goal-oriented positioning of the digital stethoscope in the future.

II. Materials and Methods

A. Single- and multi-source PCG recording

In this study, we compared single-source PCG recordings from the four auscultation areas against signals recorded from multiple points of the chest at a high spatial resolution.

To obtain high-spatial-resolution recordings, we used a custom device that we developed in our laboratory at Politecnico di Torino. This device was designed to provide the patient with a recording system he/she can use at their domicile without help from clinical staff. To achieve this goal, the device is a flexible pad based on a flexible printed circuit board mounting 48 electret condenser microphones. The microphones have a 4-mm diameter and closest neighbors are distanced by 12 mm. This results in a high-resolution inhomogeneous grid where the density of the microphones is higher over the expected traditional auscultation areas. This device should be positioned over the patient's left hemithorax as shown in Fig. 1A. Having decades of microphones covering the areas of interest tremendously decreases the need for an accurate positioning and enables the use by the patient or a caregiver. The system records a simultaneous electrocardiogram (ECG). All signals are simultaneously sampled at 1 kHz and acquired through a 16-bit ADC. More details about the hardware technical specifications can be found in [13].

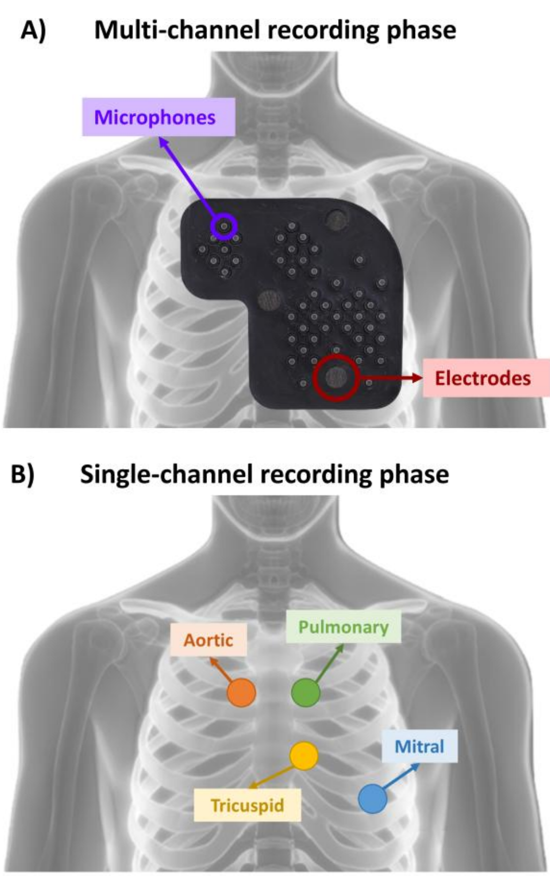

Figure 1. Recording devices and positioning over the thorax of the subject for the multi-channel (A) and the single-channel (B) recording phases.

The single-source signals were recorded using a commercial system for the acquisition of biomedical signals (ReMotus™ by IT-MeD, Torino, Italy). The system enables the recording of up to 4 signals. In our case, we first recorded the signals from the left-heart areas (mitral and aortic), along with a simultaneous ECG, then the signals from the right-heart areas (tricuspid and pulmonary). The position of the microphone probe for each area is shown in Fig. 1B. All signals were acquired at 1 kHz and digitized with a resolution of 24 bits.

B. Waveform analysis

The goal of this phase is to a) determine whether the waveform differs when the sound is recorded from different points of the chest; b) determine whether sounds recorded with different systems produce similar waveforms when recorded in a similar position. In other words, we analyzed the impact of the recording position and of the recording system on the heart sound waveforms.

For this scope, two time-averaged sound waveforms were extracted from each PCG signal, one for the first (S1) and one for the second (S2) heart sound. Averaging was performed after heartbeat segmentation, taking as reference the peak of the depolarization extracted from the simultaneous ECG. From each heartbeat, two segments were extracted:

- **S1 segment**: 300-millisecond segment starting 50 milliseconds before the R-peak;
- **S2 segment**: 300-millisecond segment starting with a delay with respect to the R-peak equal to 30% of the cardiac cycle duration.

The S1 waveform was obtained as the average of all S1 segments. Equally, the S2 waveform. The four reference waveforms (from the mitral and tricuspid areas for S1, and from the aortic and pulmonary areas for S2) were compared against the point-wise mapped waveforms. First, visual inspection was conducted to highlight the similarity of the waveform over the space and the consistency between the reference waveforms and the waveforms mapped at the corresponding positions. Then, a quantitative analysis was carried out by computing the distance between the four reference waveforms and the point-wise mapped waveforms. A correlation-based distance metric was leveraged. We expect to find the lowest distances over the expected auscultation areas.

C. CTIs estimation and objective-driven channel selection

The goal of this phase is to determine the variability of the estimate of the CTIs over the chest and to automatically select a channel providing a good enough estimate by mimicking human reasoning.

The estimation of the CTIs grounds on the correct identification of the time of closure of the four cardiac valves. We performed the identification through an envelope-based algorithm we previously developed and published [14]. The identification grounds on the Shannon Energy envelope (SEE) of the signal, which highlights the energy contribution of the two main heart sounds. The SEE, estimated using a moving window, preserves the time resolution of the signal and has a high discriminative power of the sound components (mitral and tricuspid in S1, aortic and pulmonary in S2), enabling their detection. Our previous experience shows that a minimum SNR of 10 dB is required to obtain a relative estimate error lower than 10%, which is considered sufficient for most clinical applications. The SNR varies over the chest. Also the relative importance of the two components to the recording varies over the chest. In most cases, the left-heart component is prevalent, but the relative contribution of the right-heart component changes. Our approach works best when the two components contribute similarly. In this case, two well-separated peaks appear in the SEE and their detection is robust.

The above-stated considerations, combined with more knowledge about cardiac physiology and technical constraints, allow us to define an objective function for optimizing the channel selection for CTI estimation. Instead of selecting a single channel, we divided the channels into subgroups using agglomerative hierarchical clustering and a correlation-based similarity metric [15]. Clustering helps avoiding that an outlier channel is selected and increases the robustness of the estimate. For each cluster and each valve, we computed four parameters:

- **SNR**: average Signal-to-Noise Ratio of the heart sound (S1 for mitral and tricuspid, S2 for aortic and pulmonary). It should be maximized to ensure good quality.
- **Prom**: prominence of the smallest of the two SEE peaks in the heart sound. It should be maximized to ensure the highest separability of the two components.
- **Var**: variance of the estimated time of closure within the cluster. It should be minimized to ensure the highest robustness of the estimate.
- **Resp**: Root-mean-squared error of the variation of the time of closure over time against the respiration (extracted from the ECG R-peaks amplitude). It should be minimized because the variation of the closure of the cardiac valves is expected to follow the respiratory cycle.

Each parameter was normalized by dividing by its maximum values. We defined the objective function as

$$f(c) = \frac{SNR_N(c)+prom_N(c)+(1-var_N(c))+(1-resp_N(c))}{4}. \tag{1}$$

where c is the cluster. The cluster presenting the maximum value of f(c) was selected and the corresponding estimate of the time of closure of the cardiac valve under analysis selected as final. The definition of the objective function responds to the specific requirements of our goal. A different objective function could be defined for different clinical goals.

D. Sample population and experimental protocol

We applied our analysis to a sample population of 42 volunteers. The volunteers declared no previous history of cardiopathy or cardiovascular disease. This is a requirement for this first-stage analysis because cardiovascular conditions may alter the sound waveforms. Volunteers were varied in

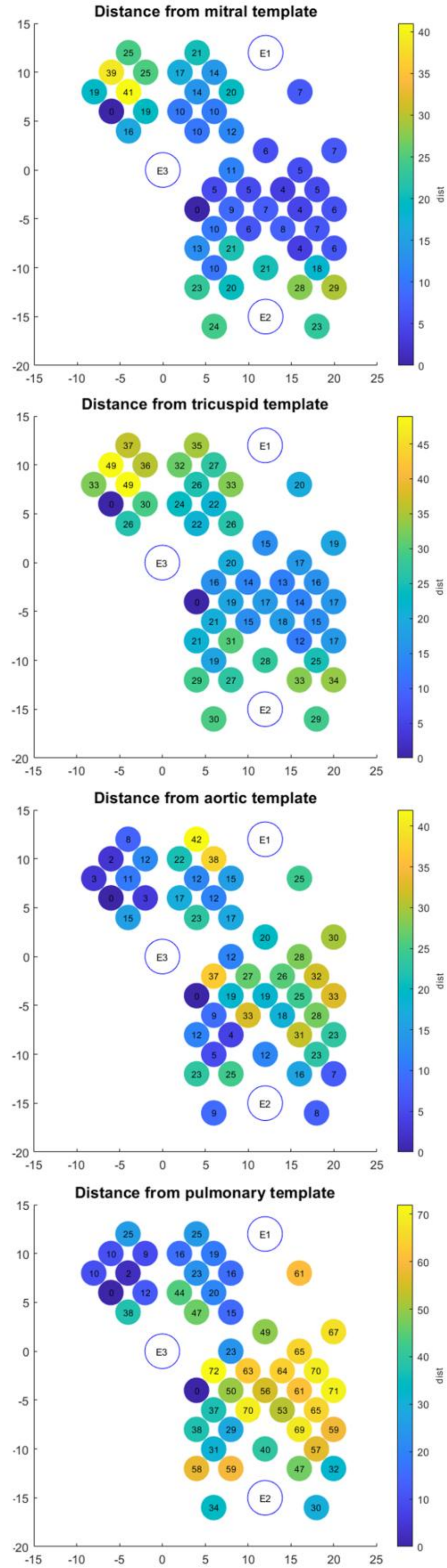

Figure 2. Maps of the distance between the heart sounds (S1 for mitral and tricuspid, S2 for aortic and pulmonary) recorded at each position of the chest against the reference auscultation area.

terms of gender (50% females) and Body-mass index (17 to 32 kg/m^2) to ensure generalizability of the results regardless of the anatomy of the thorax.

Subjects were divided into pairs. In each pair, one subject played the role of the patient, and the second played the role of the caregiver. In the first phase, the volunteer-caregiver performed the recording on the volunteer-patient using the multi-source device. Then, an investigator with expertise in digital auscultation recorded the single-source signals from the four auscultation areas. In the second phase, the two volunteers switched roles. The experimental protocol was devoted to demonstrating the usability of the device by inexperienced users, as a primary outcome.

The study was conducted in adherence with the Declaration of Helsinki and approved by the Ethical Committee of Politecnico di Torino (protocol number 16863/2021 approved on 28 May 2021).

E. Statistical analysis of the estimates

For each subject, four single-channel recordings from the four auscultation areas and one multi-channel recording at high spatial resolution were available. The time of closure of the four cardiac valves was estimated for all 48 signals from the multi-channel recordings and from the single-channel recording corresponding to the valve-specific auscultation area.

For each subject and each valve, three estimates were taken into consideration for statistical analysis:

- **Single**: The estimate obtained through single-channel auscultation at the valve-specific auscultation area;
- **Manual**: The estimate obtained through the channel manually selected by the investigator showing a good separation between the left- and right-heart components;
- **Multi**: The estimate obtained through the grioup of channels automatically selected as described in Section 2C.

A pair-wise correlation analysis was performed to determine whether the estimates were consistent. The correlation was assessed using Pearson's correlation coefficient. The confidence interval of the correlation coefficient was verified to confirm its validity. Moreover, the root-mean-squared error (RMSE) between pair-wise estimates was computed to get the measurement error.

III. Results

In the following paragraphs, the results of the above-described analysis are presented.

A. Consistency of the heart sounds waveforms

Fig. 2 shows a spatial representation of the distance between each heart sound waveform and the reference waveform recorded at the traditional auscultation area. Blue dots represent positions where the waveforms are similar to the one recorded at the corresponding auscultation area. It can be observed that blue areas are consistent with expectations according to the distribution of the microphones presented in Fig. 1. This confirms that different recording systems record similar waveforms if the position is preserved. On the contrary, the resulting high spatial variability of the distances show that sounds recorded in a different position result in a different waveform from a morphological perspective, which may impact the detection of the time of closure of the valves.

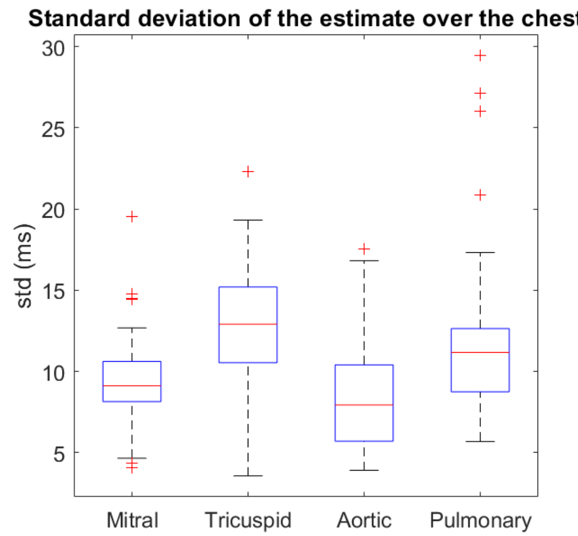

Figure 3. Boxplots of the standard deviation of the estimates of the time of closure of each cardiac valve over the chest and over the sample population.

B. Spatial variability of CTI estimates

The previous consideration is backed by results concerning the spatial variability of the estimates of the time of closure of the four cardiac valves, reported in Fig. 3. The boxplots show the standard deviation of the chest across the sample population. Except for a few cases where the standard deviation is lower than 5 milliseconds, in most cases the standard deviation is high enough to cause issues in the clinical interpretability of the derived CTIs.

C. Effect of channel selection on CTI estimates

Table I reports the correlation coefficients and the RMSE for the estimates of the time of closure of the four cardiac valves across three pair-wise comparisons: multi vs single, manual vs single, multi vs manual (according to the definitions provided in Section 2E).

TABLE I. Correlation coefficients and RMSE of the pair-wise comparison among auscultation modalities

Heart valve [a]	Correlation coefficients			RMSE (ms)		
	Multi vs Single	*Manual vs Single*	*Multi vs Manual*	*Multi vs Single*	*Manual vs Single*	*Multi vs Manual*
M	0,60	0,42	0,73	11	15	9
T	0,49	0,29	0,73	16	20	13
A	0,92	0,87	0,96	9	11	6
P	0,84	0,80	0,90	13	20	15

a. M: mitral, T: tricuspid, A: aortic, P : pulmonary

Results demonstrate that a) a human annotator would not select a signal waveform corresponding to the ones recorded

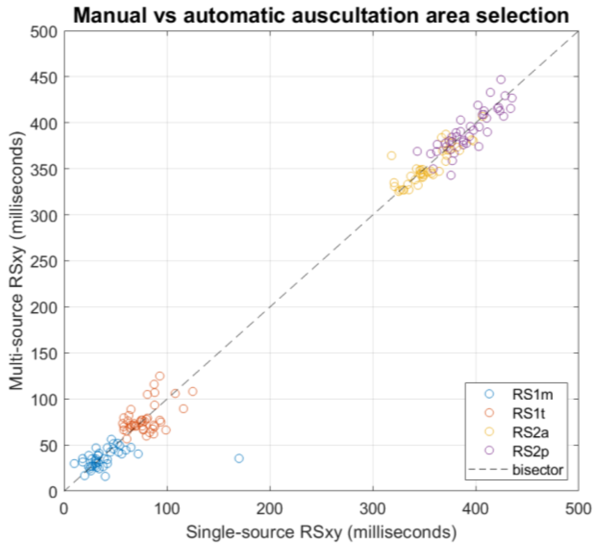

Figure 4. Correlation plot of the estimates obtained using single-source PCG (x-axis) and multi-source PCG (y-axis).

at traditional auscultation areas when it comes to discriminating left- and right-heart components; b) the defined objective function produces estimates that correlate best with the manual annotation instead of the single-source recordings.

Fig. 4 presents the correlation plot of the estimates for the multi vs manual comparison, which yields the highest correlation values and proves the suitability of the described algorithm to mimic human reasoning when deciding if a PCG signal shows good discrimination between the left- and right-heart components. The plot confirms the consistency between the estimates.

IV. Discussion

The analysis of the spatial variability of heart sounds is not common in the state of the art. Most previous works involving multi-channel recordings ground on 2 to 6 microphones, typically positioned over the traditional auscultation points. Among them, three works used multi-source recordings for signal enhancement [16], [17], [18]. Concerning clinical goals, multi-channel PCG proved valuable for coronary artery disease diagnosis [19], [20], [21], murmurs classification[22], [23], and ejection fraction estimation [24]. Paiva et al. used two microphones for CTI estimation and reported differences in the estimates [25]. Few documented examples exist of recording systems capable of achieving a similar spatial resolution compared to the system leveraged in this study [26], [27]. To our best knowledge, no previous study analyzed the variation of the heart sound waveforms and the resulting CTIs over the chest with a high spatial resolution.

According to our results, we can first confirm that, even with naïve instructions, inexperienced users are capable of performing a reliable positioning of the multi-sensor device. This was confirmed by visual inspection and backed by a quantitative assessment of the correlation between the multi-channel recordings and the reference recordings over the four auscultation areas: the most similar waveforms were found where expected for the population of interest. In case the goal of digital auscultation is consistent with the goal of traditional auscultation, the channel can be easily selected within the areas of interest. This applies, for example, for the automatic detection of murmurs, for which the traditional auscultation areas represent a valuable gold standard.

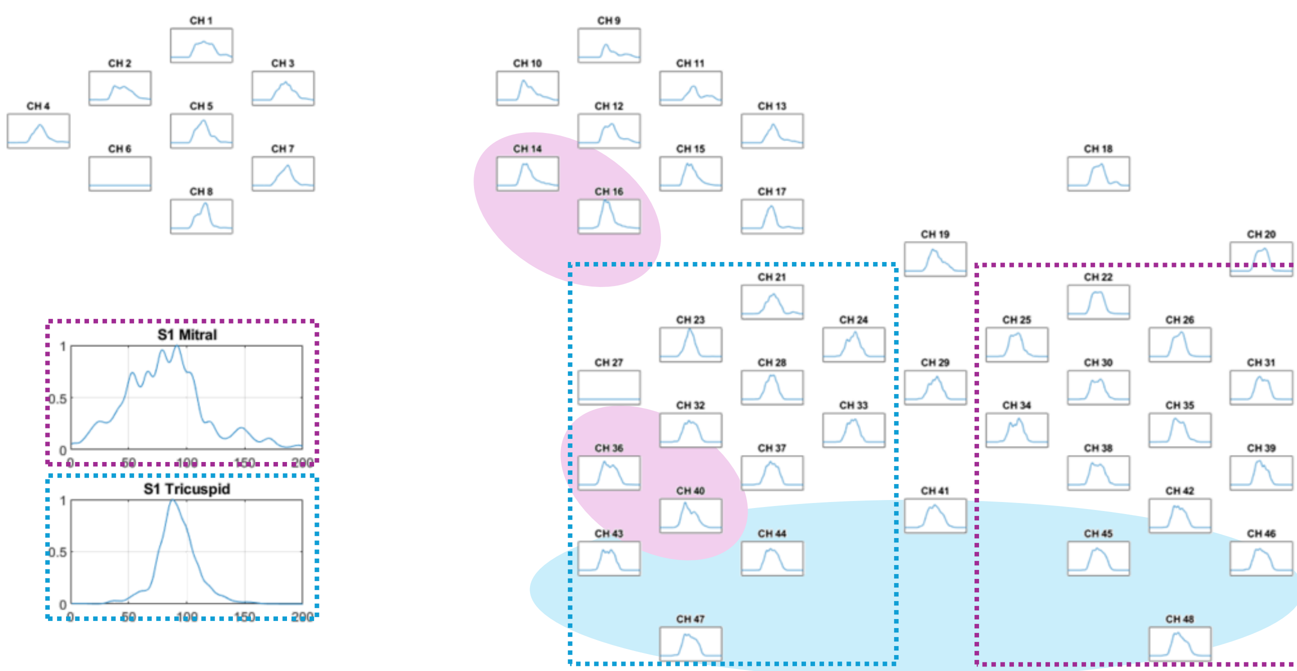

Figure 5. Spatial analysis of the average envelope waveform for S1. The dotted boxes represent the channels roughly corresponding to respectively the mitral (in purple) and tricuspid (in blue) auscultation areas. It can be appreciated that the single-source waveform shape can be found within the dotted areas. Colored ovals represent the auscultation areas picked by the algorithm for the estimation of the time of closure of respectively the mitral (in purple) and the tricuspid (blue) valves.

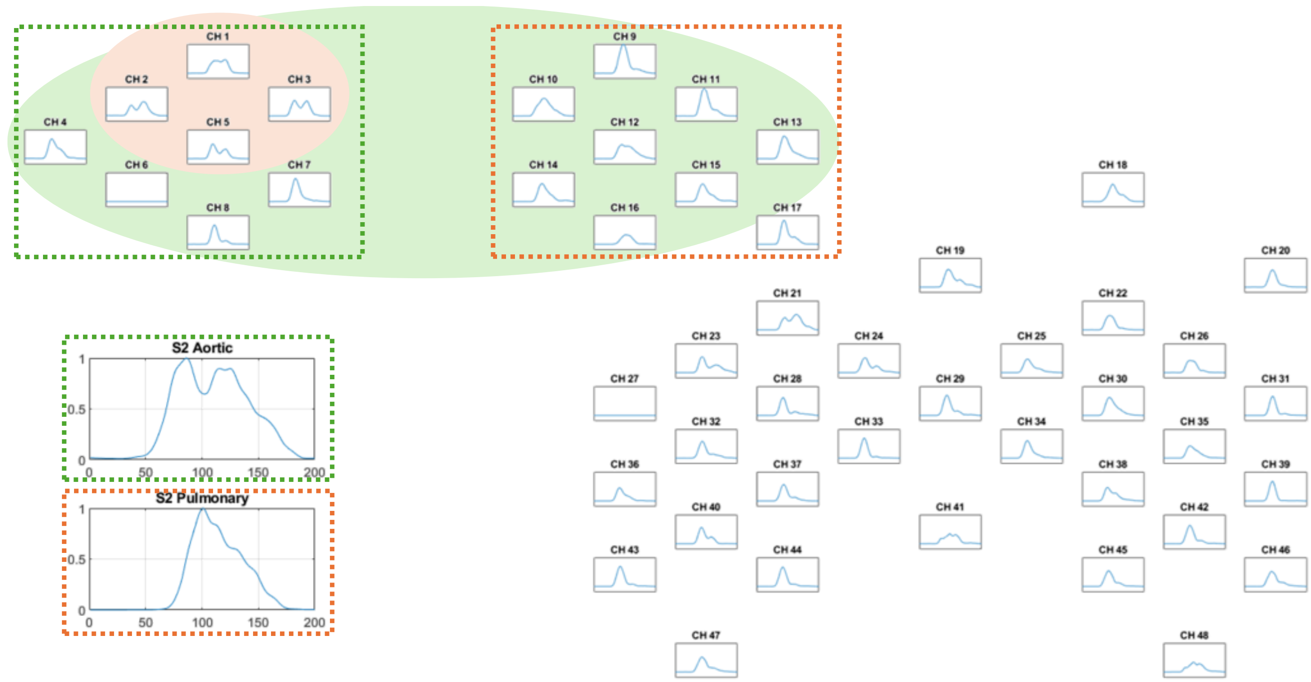

Figure 6. Spatial analysis of the average envelope waveform for S2. The dotted boxes represent the channels roughly corresponding to respectively the aortic (in green) and pulmonary (in orange) auscultation areas. It can be appreciated that the single-source waveform shape can be found within the dotted areas. Colored ovals represent the auscultation areas picked by the algorithm for the estimation of the time of closure of respectively the aortic (in green) and the pulmonary (in orange) valves.

On the other hand, our results show that a different channel selection may be most suitable for different goals. When it comes to estimating the time of closure of the cardiac valves for electromechanical assessment purposes, our variability analysis shows that a standard deviation between 8 milliseconds and 13 milliseconds can be expected when using signals from different points of the chest. This is consistent with the findings of previous works dealing with the spatial variations of CTIs [28]. According to acoustics and the velocity of propagation of acoustic waves through biological tissues, the reported variability is not due to a physical phenomenon but reflects errors in the estimate. Errors may arise from a variety of reasons and depend on the estimation method.

In this work, we tested an energy-efficient envelope-based estimation method. The method is particularly suitable for edge computing and is therefore of interest for its implementation on board the recording device. The method grounds on the separability of the two inner components of the two heart sounds in a suitable domain: therefore, it works the best when the two components appear to contribute to the sound almost equally. In this case, an expert user, when asked to perform a single-source recording, would not follow the traditional auscultation areas but would pick a signal where the two components can be distinguished at visual inspection. Please, note that the components cannot be distinguished at listening in any case because of the limitations in bandwidth of the human ear.

Fig. 5 and 6 shows an example of the average SEE waveforms for a random subject from the sample population, respectively for the first and second heart sounds. It can be appreciated that even if the templates are representative of the waveforms in the corresponding areas, those are not necessarily the ones where the two components are best visible (i.e., there are two clear peaks in the envelope). Instead, the algorithm is capable of automatically selecting a subset of signals where the separability of the two components is maximized, thanks to the definition of an appropriate objective function. The objective function can obviously be adapted to solve different clinical problems.

To our best knowledge, this study is the first proposing an insight into the spatial variability of the heart sound waveforms. The availability of a recording system for multi-channel sound recording at high spatial resolution opens to a thorough optimization of the auscultation area for different clinical goals. This study presents some intrinsic limitations that will be faced in future works. On one side, the choice of presenting an analysis on healthy volunteers limits the generalizability of the findings to the patient population, where the spatial variability of the sounds may differ. Moreover, a thorough validation of the estimated intervals against a clinically acceptable gold standard is missing and will be carried out in the future to further confirm the presented findings.

V. Conclusion

In this study, we analyzed the variation of the heart sound waveforms in different recording points widely distributed over the patient's chest with a high spatial resolution. We found that, even if the waveforms were consistent to what recorded with a traditional system over the traditional auscultation areas, the reference waveforms were not optimal for discriminating the left- and right-heart components withing heart sounds and obtain reliable estimates of the CTIs. We defined an objective function mimicking human reasoning for selecting a group of channels to be used for the estimate and found good correlation with a human annotator. Our analysis moves the first steps into the spatial analysis of heart sounds, which could help clinicians and researchers in the future in defining objective-oriented recording points according to the application of interest.

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