Development of a System for the Automatic Detection of Air Embolism Using a Precordial Doppler

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***Abstract*— Venous air embolism (VAE) is the air bubble accumulation in the right side of the heart. Changes in Doppler heart sound (DHS) are characteristic of VAE, and the anesthesiologist has to pay attention to this event continuously, which may not always be possible. This work aims to study different features of the heart sound through a precordial Doppler, that may provide useful information on VAE episodes.**

A clinical protocol was designed, and DHS was collected at baseline and following infusions of saline with 4 distinct volumes (1ml, 5ml, 8ml and 10ml), and two infusion rates (slow and fast), given by central and peripheric catheters. Signal was pre-processed, the envelope of each signal was extracted, and five features were implemented and evaluated: frequency corresponding to 95% of Welch periodogram power (f95), frequency corresponding to 50% of Welch periodogram power (f50), frequency corresponding to maximum power spectral density (fm), entropy (E), and frequency corresponding to maximum energy of a wavelet transform (freqwav). Relation between extracted features and saline infusions were studied and compared to baseline values. A Graphical User Interface (GUI) with a database of Doppler heart sounds and annotations was also developed.

Although features present a high variability between patients, E presents a better performance showing an increase in response to saline injections (in 75% injections), followed by f95 (62%), fm (56.3%), freqwav (37.5%) and f50 (0%).

I. INTRODUCTION

Venous air embolism (VAE) is a serious complication that may occur during surgical procedures [1]. It is the result of air bubble accumulation in the heart, causing over distension of the right side, and when not detected in time, may lead to many health problems such as cyanosis, stroke, obstruction of pulmonary blood flow, or cardiovascular collapse [2, 3].

The accumulation of air bubbles is caused by the existence of pressure gradients between the right side of the heart and the incisional area, and it is more frequent for the sitting position (incidence of 25%), but may also occur in the prone, supine, or lateral positions (incidence ranging between 15% and 25%) [2].

Both volume and rate of air accumulation are dependent on the size of the vascular lumen, and the pressure gradient. These factors determine the morbidity and mortality of any episode of VAE, as well as the positioning of the patient [3, 4]. Neurosurgeries have a high risk of VAE episodes since the pressure gradient of the head is negative when compared with the atmospheric pressure, and the probability of air entry in the blood flow is higher.

Pathophysiologic pathways are highly dependent on the volume of air accumulated within the right ventricle. If the rate of embolism is high (approximately 5ml/kg), the probability of air bubble accumulation in the heart is higher. As consequence, the normal blood flow is interrupted. This outflow obstruction may be the result from the inability to decompress the tension of ventricular wall, which leads to heart failure and cardiovascular collapse. When the volumes of VAE are moderate, a decrease in cardiac output, hypotension, myocardial and cerebral ischemia, and even death, may occur [5].

To detect and prevent these problems, monitoring protocols are implemented, including a precordial ultrasonic Doppler probe, the insertion of a right atrium line for air aspiration, and the institution of high flow oxygen [2]. The Doppler probe allows a real-time monitoring of the heart sound [5]. Changes in heart tones, usually referred to as "mill-wheel murmurs" are characteristic of VAE and the anesthesiologist has an important role in the detection of these characteristic sounds, and subsequent intervention [2].

At this point, detecting VAE relies on the continuous attention of the anesthesiologist throughout surgical procedures to cardiovascular changes, and the Doppler sound beat-to-beat. Since the anesthesiologist needs to focus his attention on other tasks, detection of VAE may be disregarded; especially with the occurrence of false positives.

In this context, it is imperative to assist the decision of the anesthesiologist by studying and developing other robust methods of VAE detection. The study of the Doppler heart sound (DHS) at different levels may provide valuable information for VAE episodes detection.

The main goals of this work were to study DHS changes, propose different methods for its processing and features extraction; evaluate the proposed metrics in response to different administered volumes and administration rates of intravenous saline injections; and to develop a system for offline analysis of the DHS, with features extraction and visualization.

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I. MATERIALS AND METHODS

To perform data acquisition, a clinical protocol was designed and submitted for approval to the Department of Education, Training and Research, and to the Ethics Committee of Santo António's Hospital, Centro Hospitalar do Porto. After approval, data collection was initiated.

During intraoperative management, the precordial Doppler was placed on the patient's chest according to manufacturer recommendations [2, 4, 5]. All patients were submitted to a total intravenous anesthesia (TIVA) of propofol and remifentanil [6]. To study which features allowed a better discrimination between normal blood flow and turbulent blood flow, the designed clinical protocol included a series of saline injections through peripheral and central catheters, at different rates and volumes of infusion, as a way to study how these conditions affect changes in the Doppler sound. The protocol was designed in this way because injection of air bubbles was not viable, and the occurrence of VAE is low for the available time span. The administrations of saline started after the patient was stabilized following induction. In total, each patient received 16 saline injections (with intervals of 15 seconds) with volumes of 1ml, 5ml, 8ml, and 10ml at two distinct administration rates (one as slowly as possible, and the other a fast injection) by peripheral and central catheters.

Data was collected using the RugloopII Waves© (Demed, Belgium). Besides data from the monitors, age, weight, height, and gender were registered. DHS was collected at 8kHz, and all injections moments were annotated by the researcher for the following analysis. MATLAB 2012 was used for the analysis and graphical interface development.

A. Heart Sound Pre-Processing and Segmentation

Pre-processing of the signal included filtering the unwanted frequency components. A Butterworth low pass filter was applied to the DHS with a cut-off frequency of 900Hz in both directions to avoid phase distortion. The Savitzky-Golay filter [8] was also implemented, because it smoothes the DHS [9]. To apply this filter to the DHS, a polynomial of order 3 was chosen, with a frame size of 21, since higher frame sizes highly distorted the signal. These filters were applied to the DHS with different purposes: the first to remove frequency components outside the band of interest, and later extract signal features for analysis; and the second to provide a smoothed version of the signal, and construct the envelope for heart cycle identification. Figure 1 shows the application of the two filters to a frame of DHS.

The DHSs were down-sampled to 2kHz and normalized. Frames of 3 seconds of DHS were analyzed, since this is the adequate time to detect at least one complete cardiac cycle [7] (afterwards, this frames will be useful to make the cross correlation and estimate the heart rate of each patient).

Following, the signal envelope was extracted (an example may be seen in Figure 2), and the signals segmented into heart cycles. The energy of the signal was calculated using the Savitzky-Golay filtered DHS in windows of 0.5s (duration of one cardiac event). Then peaks of the envelope were detected using a peak-picking function with a minimum peak distance of 0.1s. The detected peaks were then interpolated with a cubic Hermite spline to provide a smooth envelope of the signal.

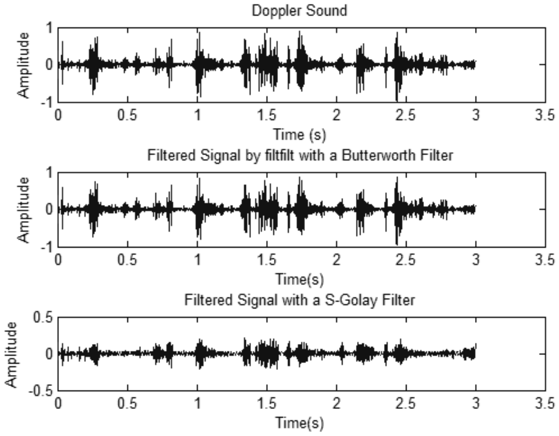


Figure 1. Original signal and filtered signal using Butterworth and Savitzky- Golay filters.

After obtaining the envelope, the cross-correlation was calculated for two consecutive frames, and local maximums were detected, corresponding to the heart rate estimation.

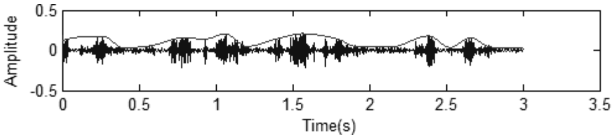


Figure 2. Envelope of one frame of DHS.

B. Doppler Heart Sound Features

Frequency content of the DHS was analyzed by estimating the power spectral density (PSD) of the signal through the Welch method. The PSD Welch was obtained using a 150 points window with 50% overlap, and a Blackman window.

After obtaining the PSD for each heart cycle, three features were extracted: the frequency corresponding to the limit of 95% of PSD power (f95), the frequency corresponding to 50% of the PSD power (f50), and the frequency corresponding to the maximum power peak of the PSD (fm). The method was applied to the signal using a moving window of 0.1s and the vectors stored.

Continuous wavelet transform (CWT) analysis was also used to study the DHS. The scales used were between 1 and 10 at a distance of 0.1. The chosen wavelet was the "Mexican Hat Wavelet" since it has demonstrated to provide good results in the analysis of the Doppler sound [10]. CWT was applied to intervals of 0.1s of the signal to extract the respective scales. With the coefficients obtained, the points of maximum energy of the signal and the respective scales were extracted, identifying the frequency band with higher energy (freqwav), which was also stored. Again, a vector with the average of this feature for the corresponding heartbeats of two seconds of each injection was created.

An additional feature proposed in this study was the Shannon Entropy (E), implemented at each segment of 0.1s, and the values with the mean of this feature for the heartbeats of the two seconds after each infusion stored in a vector.

C. Features Evaluation and Graphical User Interface

After extracting all features and store them in a text file, these were exported to Excel (Office 2010) for the remaining analysis. The features average value for each injection and patient were calculated, as well as the ratio between features values following injection and at baseline conditions (injection/baseline) for different administration rates and administration catheter.

To allow automatic extraction of the features from each signal, a graphical user interface (GUI) was developed. The GUI allows the user to choose the signal to analyze, displaying patient's demographic data, estimated heart rate, and features' values at baseline, as well as the display of the features series for the collected DHS.

II. RESULTS AND DISCUSSION

A. Data Collection and Pre-Processing

13 patients were enrolled in this study but only 10 acquisitions were afterwards studied due to a technical fault during the infusions of saline. 8 female patients (means: age 57 years; height 163 cm; heart rate 78 beats/min; and weight 67 kg).

The smooth envelope allowed an accurate estimate of the heart rate, and the identification of peaks relative to cardiac events (two for heartbeat), useful to extract the proposed features following saline injections.

B. Features Evaluation

To evaluate turbulence episodes, the average value of the features extracted from the DHS events in the 2s following injection were analyzed, for each injection, for each patient: Mean_Mf95, Mean_Mf50, Mean_Mfm, Mean_ME and Mean_Mfreqwav. The features were normalized by the baseline value, meaning the ratio between features post-injection and baseline features. The normalized values were related with the infusion volumes in terms of rates and administration catheter, for each feature. Table 1 presents the number of detectable infusions (average values for all patients) for each catheter and infusion rate.

We observed that the extracted features' values were not influenced by the infusion volumes. The different volumes only influenced the infusion time. For this reason, the number of infusion detections for each feature were analyzed, considering administration catheter (central and peripheric) and injection rates (fast and slow). A detectable change was defined as an increase above the baseline feature value, for each patient.

In general, infusions made by central catheter were more easily detected (21 positive detections) than by peripheric catheter (16 positive detections). This was expected since infusions via central catheter are given directly in the heart, where the Doppler probe is placed. Surprisingly, slow infusions were detected in more cases (21), then fast infusions (16).

TABLE I. NUMBER OF DETECTABLE INFUSIONS FOR EACH FEATURE RELATIVE TO CATHETER AND INFUSION RATE (N=NUMBER)

Catheter	**Total N infusions**	**N of detectable infusions**				
		f95	***f50***	***fm***	***E***	***freqwav***
Central	8	6	0	5	7	3
Peripheric	8	4	0	4	5	3
Infusion Rate	**Total N infusions**	**N of detectable infusions**				
		f95	***f50***	***fm***	***E***	***freqwav***
Fast	8	5	0	3	5	3
Slow	8	5	0	6	7	3

Relative to administration catheter and infusion rates, E was the feature with more infusion episodes detected, followed by f95, fm and freqwav. f50 did not present any detections. This occurred because the baseline and post-infusion values are very similar, and when average for this feature was calculated, no detections were observed.

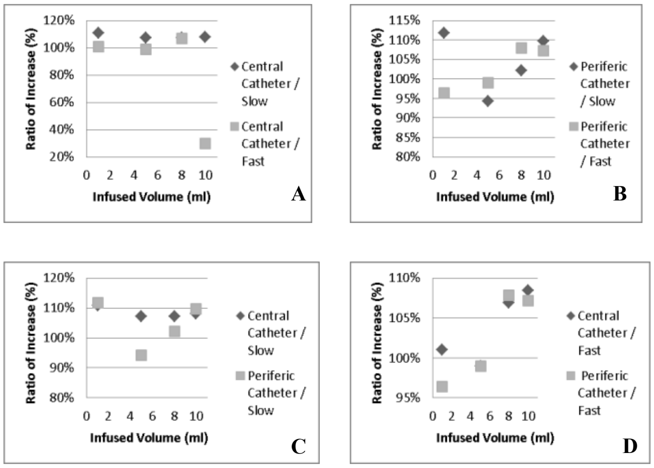
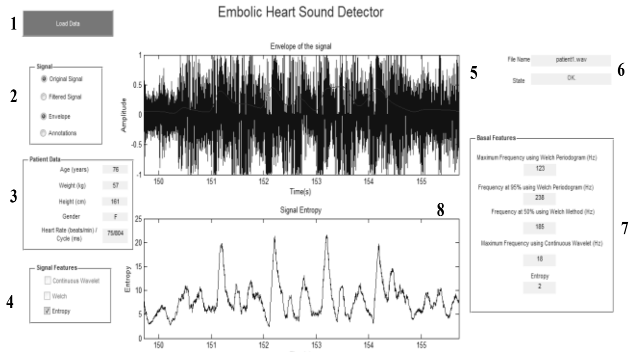


Figure 3. Relation between the infusion volume and the ratio of increase (%) of Entropy for (A) slow and fast rates of saline infusion via central catheter; (B) slow and fast rates of saline infusion via periferic catheter; (C) slow infusion of saline via central and periferic catheter; (D) fast infusion of saline via central and periferic catheter

Figure 3 presents the average results of the study sample for the Entropy feature.

The patients enrolled in this study needed to fulfill the study requirements which limited the number of patients. However, it is visible that the turbulence episodes are more evident with the increase of volume infusions, and more easily detectable when the infusions are given via central catheter. Although, all features seem to stabilize with higher infusion volumes. This could be empirically explained because there is a ceil volume from which the turbulence is maximum. So, even if the volumes administered are increased, the changes in DHS would only depend on the limit of infusion administration rate.



● Original Signal
○ Filtered Signal
● Envelope
○ Annotations

Age (years) 76
Weight (kg) 57
Height (cm) 161
Gender F
Heart Rate (beats/min) / Cycle (ms) 75/804

☐ Continuous Wavelet
☐ Welch
☑ Entropy

File Name patient1.wav
State OK

Maximum Frequency using Welch Periodogram (Hz) 123
Frequency at 95% using Welch Periodogram (Hz) 238
Frequency at 50% using Welch Method (Hz) 185
Maximum Frequency using Continuous Wavelet (Hz) 18
Entropy 2

Figure 4. Graphical User Interface and respective functions: 1.Load the wave file; 2. Select pre-processing methods; 3. Patient data relative to the *.wav file selected; 4. Select signal feature; 5. Graphical representation of the pre-processing method (s) selected; 6. File name and program state; 7. Baseline features of selected file; 8. Graphical representation of selected features.

C. Graphical User Interface

The development of an interface was useful to help the user understand how different approaches affect the signal (filtering or envelope extraction e.g.) and to see the features extracted, and correlate them with the original signal. The main result of this interface was to make available a database of Doppler sounds, standardized, with annotations of different volumes and rate of saline injections. This interface allows observing the behavior of different features for each signal, as well as the demographic data of each patient. The developed tool is presented in Figure 4, and may be adapted for on-line analysis of the DHS.

CONCLUSION

This is, to our knowledge, the first study in human patients to search for automatic methods to improve the detection of VAE episodes using a precordial Doppler, simulating DHS turbulence, although a few attempts have been made in animals [7].

Pre-processing of the signal to eliminate out of band noise and normalization was implemented; a new method for signal envelope retrieval, and heart rate estimation, based on a Savitzky-Golay smoothed DHS was applied with success to the DHS segmentation; finally, detected DHS events were analyzed for each patient, extracting several time-frequency features for analysis.

Five features were proposed in this study and evaluated in response to saline injections. There was a high variability in the results because the number of patients was reduced. The feature presenting the best turbulence detection results was the entropy since almost all saline infusions caused changes in Doppler sound, detectable with this feature, when compared with baseline DHS.

In a general way, it is possible to conclude that detection of turbulence is more evident when the infusions of saline are administered by central catheter, probably because the infusion enters directly in the heart, where the Doppler probe is placed (externally). Since we could not produce air embolism episodes, we studied episodes that produce changes in Doppler sound and were innocuous for the patient. The next step will be to acquire Doppler sound when VAE episodes occur and relate it with these results to find similarities.

Future work will include: further data collection to increase the number of acquisitions, and improve the available annotated database; the acquisition and study of heart sounds when VAE events occur naturally, to evaluate the robustness of the features in detecting air bubbles in the circulation; and the adaptation of the GUI for online analysis. This would allow to implement an advisory system that could aid the anesthesiologist detect VAE episodes more rapidly.

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