Embedding a Multichannel Environmental Noise Cancellation Algorithm into an Electronic Stethoscope

D. Della Giustina, M. Riva, F. Belloni, M. Malcangi

Abstract**—**The paper describes a multichannel adaptive algorithm for the enhancement of cardiac sounds with respect to environmental noise. It combines sounds acquired from a couple of microphones to reconstruct the transfer function of a stethoscope head and its interaction with the patient's body.

This identification process allows to perform a distortion-less noise reduction. The filter is embedded into an electronic stethoscope, composed of a traditional acoustic head and an electronic section. This instrument allows to show on a display the heart sound and to store the acquisition into a removable media transferring data to a PC.

A software tool able to reproduce, visualize, store and analyze cardiac sounds, for performing assisted diagnoses of cardiac diseases, completes the system.

A demonstrator of the tool has been realized. Experimental results show significant improvements in noise reduction, when the filtering algorithm is applied.

Keywords**—** Cardiac Sounds, Adaptive filters, Biomedical signal analysis, Noise cancellation, System identification.

I. Introduction

IN the last century, cardiovascular illnesses are the first death causes in developed countries [1]. For this reason, many efforts have been made in order to develop sophisticated techniques for the early diagnoses of cardiac disorders.

The massive diffusion of such techniques has led physicians to a progressive relinquishment of the traditional stethoscope, with a series of major consequences:

- over-crowding of clinics and long waits for examinations;
- increasing in the National Health Service costs;
- not always justified recourse to time-consuming and expensive diagnostic procedure.

This scenario was also supported by some well known disadvantages of the traditional acoustic stethoscope:

- its performances are strictly linked to the audio capability of the user;
- high levels of environmental noise invalidate auscultations;
- analytic comparisons among delayed auscultations are impossible.

Modern electronic stethoscopes [2]-[6], already available on the market, take advantage of modern technologies to reduce hese disadvantages. They allow to adjust the volume of auscultations and to store them on a non volatile memory (e.g. on a PC).

Top level devices are mechanically designed to reject most of the environmental noise, even though its electronic cancellation is not usually implemented.

Commercial software tools [7]-[8] for the record, display and replay of acquired sounds are available too.

This paper presents a comprehensive platform [9]-[10] which includes both hardware and software and allows to acquire, to organize and to analyze cardiac signals. Fig.1 shows a block scheme of the whole system.

The core of the proposed solution is a new digital stethoscope, performing an environmental noise reduction via a distortion-less electronic filter.

The digital stethoscope is composed of:

- a low noise signal detection system - Acquisition Board (AB) - that acquires the low level heart sound by means of two high sensitive microphones, and adapt it for the analog to digital conversion;
- a processing unit - Processing Board (PB) - which combines the two signals and reduces the noise using a digital adaptive algorithm, described in the next Section. It also contains hardware resources to display and play in real-time the filtered heart sound.

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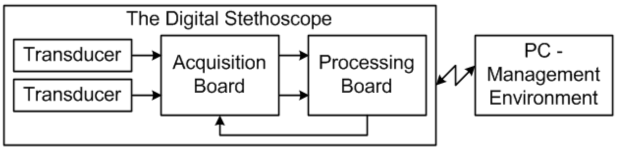

Fig. 3 Block scheme of the system.

The AB is placed close to the stethoscope head, while processing section is included in a portable device. In the PB, data are written to a Secure Digital (SD) card and therefore they can be easily copied to a PC, where a software framework stores auscultations into a relational database and allows further consultations.

The paper is organized as follow: Section 2 describes the noise reduction algorithm and the hardware, explaining the limits of traditional approaches. It also introduces the software platform for management and analysis of auscultations. Experimental results are reported in the Section 3 and confirm the effectiveness of the proposed solution.

II. Methods

A. Noise Reduction Technique

One of the most distinctive aspects of the proposed system is its ability to reduce background noise. Several approaches have been proposed in the literature and different solutions were riddled. Some considerations show that many traditional techniques cannot be applied.

The frequency range of the cardiac signal is limited below 1 kHz [11], as shown in Fig.2.

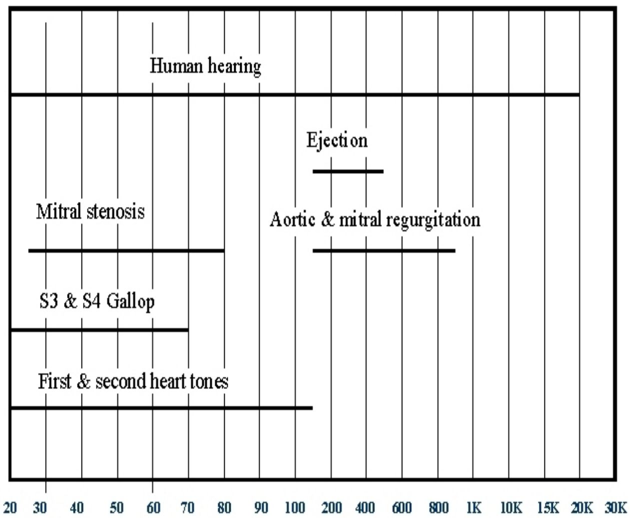

Fig.2 Frequency range of the main cardiac sounds.

The frequency band of the background noise always overlaps this range. To enhance the heart beat x_h with respect to the noise x_n, traditional stethoscopes employ their heads that act as a band–pass filter h_b on the frequency range of interest. Denoting the convolution product by ×, the collected sound is $y = x_h \times h_b + x_n \times h_b$, where the first term of the sum is the actual sound, while the second one is the noise. This process is depicted in Fig.3.

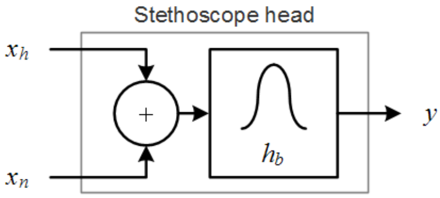

Fig.3 Noise filtering scheme of a traditional stethoscope.

The sound is usable for diagnostic purposes only if the noise is small enough with respect to the heart sound ($\|x_n \times h_b\| << \|x_h \times h_b\|$). If this condition is not satisfied, filtering is required. It is worth underlying that physicians are trained to diagnose sounds from traditional stethoscopes so, any further modification of the frequency spectrum, besides those introduced by h_b, can lead to a wrong diagnosis.

The most naive approach to achieve noise reduction should employ a further low-pass electronic filter h_{lp} on the sound $(x_h + x_n) \times h_b$ (Fig.4).

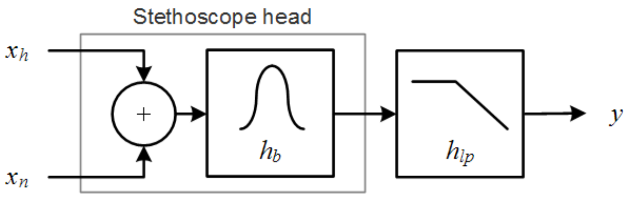

Fig.4 Noise filtering scheme of the elementary solution.

This solution would work if most part of the energy of the background noise was associated to high frequencies. Optimum filtering techniques have been considered too (Fig.5).

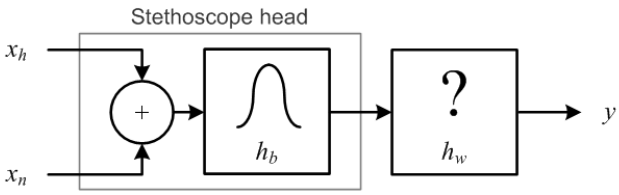

Fig.5 Generic optimum noise filtering scheme.

These algorithms move the noise reduction problem to a dynamic identification problem. Among these latter methods, standard Wiener filtering technique [12] based on a single-channel acquisition is not applicable here, because the noise reduction is strictly linked with a signal distortion that makes impossible a traditional diagnosis. In fact, let h_w be the impulse response of the optimum Wiener filter, the output signal is $y = h_w \times (x_h + x_n) \times h_b$ which is different from the expected signal $x_h \times h_b$. In principle, distortions could be reduced using a multiple-channel algorithm [13]. Nevertheless, in both cases (single and multiple channels) noise has to be estimated in the operative conditions (i.e. when the stethoscope head is applied on the patient's body) without detecting the heart beat. Evidently, such a condition cannot be satisfied. So, standard and multi-channel Wiener filtering methods cannot be successfully applied.

The approach proposed in the paper uses a stethoscope head equipped with two small pipes. While the first pipe is directly connected to the stethoscope head, the latter is put just beside, as shown in Fig.6.

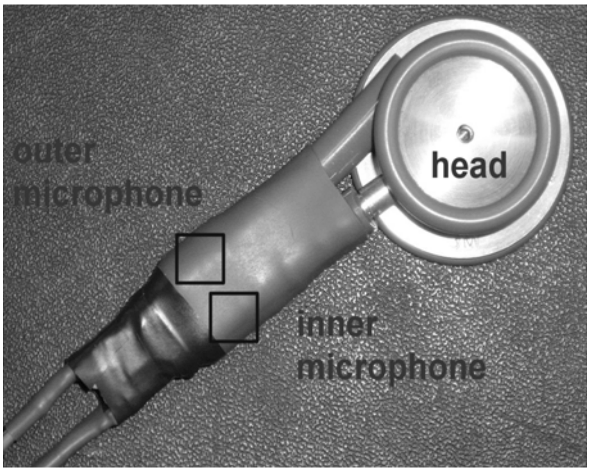

Fig.6 Stethoscope head equipped with the two microphones.

A microphone is placed inside of each pipe. Inner microphone perceives the sound from the inside of the head, i.e. both the heart beat and the noise. Outer microphone detects the noise only, due to the very low level of the cardiac signal (hypothesis experimentally verified). The pipes have the same length and are place close enough so that the spatial dependencies of the sound waves are negligible in the frequency range of interest.

The filtered signal cannot be easily yielded through a direct subtraction (Fig.7), because the two transducers do not sense the noise in the same way.

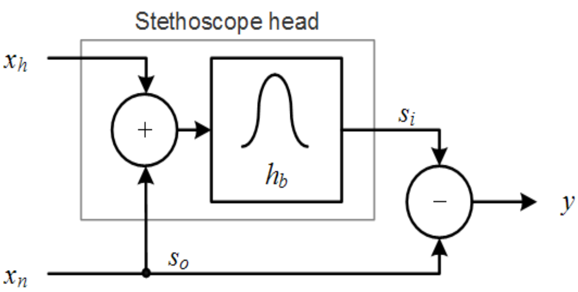

Fig.7 Direct difference method.

Differences between the signals, caused by the stethoscope head and by the interaction with the patient body, must be considered before subtracting (Fig.8). The key point of the described approach is the best estimation of the physical system response h_{wd} .

Its time-variant dynamics depends on:

- type of stethoscope head;
- patient's build and posture;
- position and pressure of the head over the patient's body;
- possible presence of clothes.

So, an adaptive approach must be adopted. In principle, it consists in minimizing the following functional:

$$\mathrm{E}(h_{wd}) := \left\| s_i - s_o \times h_{wd} \right\|^2 \tag{1}$$

where s_i and s_o are the signals from the inner and outer microphones respectively.

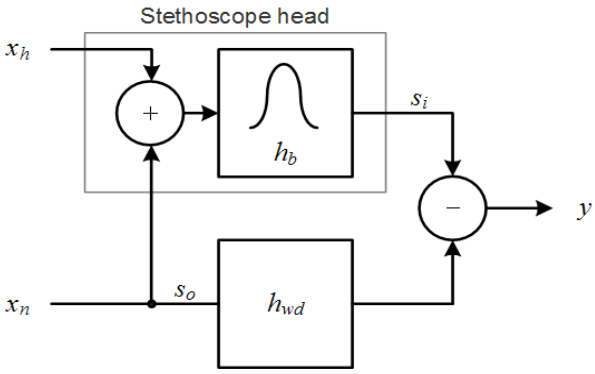

Fig.8. Weighted difference method.

Its numerical implementation can be derived by optimizing the quadratic form (LMS algorithm):

$$\mathrm{E}(h_{wd}) = \sum_{k=1}^{m} \left[s_i(k) - \sum_j s_o(k-j) h_{wd}(j) \right]^2 + \sum_{k=1}^{n} a(k) h_{wd}^2(k) \tag{2}$$

where n is the number of coefficients of the filter, m is the number of the processed incoming samples and $a(k) > 0$, for all k. The first term in (2) is the numerical translation of (1), while the second term takes into account the energy of the filter itself. Functional (2) admits a unique minimum, that corresponds to the solution of the linear system $h_{wd} \cdot A = B$, where

$$A_{ij} := \sum_{k=1}^{m} s_o(k-j) s_o(k-i) + a(i)\delta(j-i)$$

$$B_{1i} = \sum_{k=1}^{m} s_i(k) s_o(k-i)$$

for $i, j = [1, n]$.

Theoretically, h_{wd} should be updated for each new incoming ADC sample. However, since the dynamic variations of the system are quite slow compared to the sampling frequency, the filter can be calculated only few times per second. This assures a good compliance between the estimated system and the real one, without an excessive computational burden.

Once the target transfer function is correctly estimated, subtraction between the two signals can be carried out without introducing any distortions.

B. Hardware Description

A demonstrator of the stethoscope, implementing the noise-reduction technique presented in the previous Section, has been realized according to the general description reported in Fig.1. A detailed description of each part is presented in the following.

1) Transducers

The first stage of the data acquisition chain consists in a couple of sensors. High sensitive microphones, with flat frequency response in audio range, have to be selected. The devices have to be also small and light not to limit the portability of the instrument. Electret Condenser Microphones (ECM), which completely fulfills the above specifications, have been employed. Moreover, such solution does not require power supply for microphone.

2) Acquisition Board (AB)

The core of the AB is an analog front-end electronics, which adapts the signals received from the microphones for the AD conversion. According to the block scheme in Fig.9, five main subsystems can be highlighted:

- preamplifier: amplifies the signals to make them less sensitive to electronic noise;
- programmable-gain amplifier (PGA): adapts signal amplitude to the ADC dynamic range also included in AB. Gains G are automatically set by PB via a shift register;
- anti-aliasing filter: cuts off frequencies higher than 2kHz (out of the heart beat spectrum) to avoid frequency fold-back;
- single-ended to differential signal adapter: provides a differential pair to the ADC input;
- feedback network: eliminates the random DC offset voltage of the microphones.

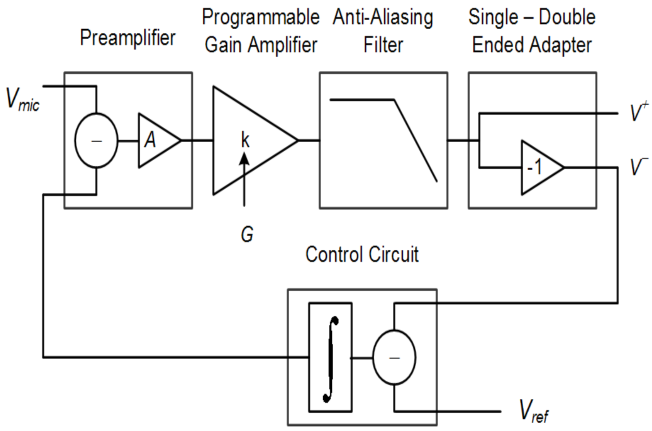

Fig.9 Block scheme of the analog front-end in the AB.

Since the actual signal is derived from the difference between two waveforms of comparable amplitude, a very high signal to quantization noise ratio (SQNR) is required. Therefore, a 4th order, 24 bits $\Delta-\Sigma$ modulator with a sampling frequency of 40 kHz has been selected.

In the present implementation, AB is connected to PB via Ethernet cables. One of them transmits the outputs from the ADC to the PB and the control commands from PB to ADC. Differential transmission is used in order to limit the external and cross interferences. A second cable is used to supply power the AB from the PB, and to manage the shift register used to program the PGAs. To reduce the contribution of the power supply noise, power lines from PB are locally regulated by a low-dropout stage

3) Processing Board (PB)

The PB, whose block scheme is reported in Fig.10, contains two main digital units: an FPGA and a DSP. The first logic unit coordinates the data transfer from the AB to the PB. Data from ADCs are filtered and stored, so that the DSP can perform the noise reduction algorithm.

The clean heartbeat is then distributed to:

- a DAC, which allows the user to listen to the sound, via common headphones;
- a display, where the sequence is plotted;
- a removable storage media (SD card).

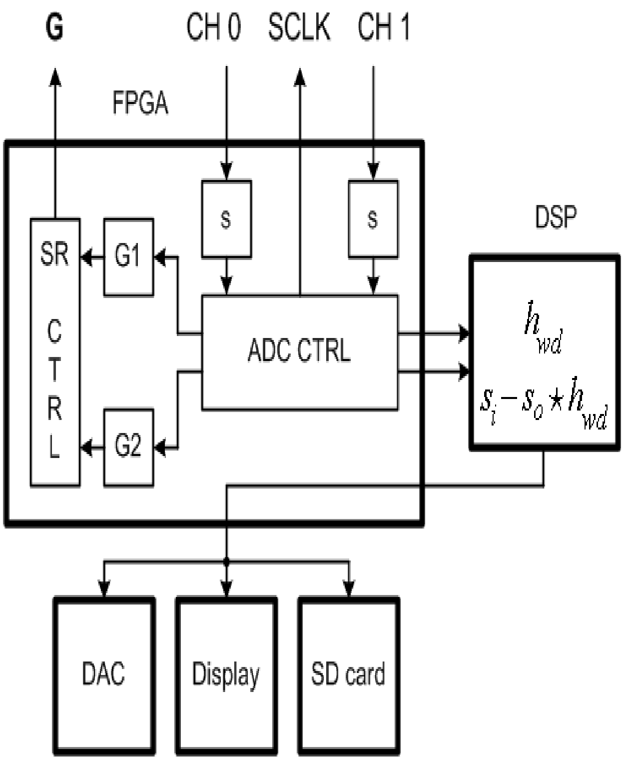

Fig.10 Block scheme of the digital section of the PB.

The DSP has been chosen taking into account that the computational complexity of the algorithm is the sum of these contributes:

- $O(n^2\, m/2)$ for the calculation of A;
- $O(n\, m)$ for the calculation of B;
- $O(n^3)$ for matrix inversion;
- $O(n^2)$ for matrix multiplication;
- $O(n\, m)$ for the application of the filter.

The optimum tradeoff between precision and computational time has been obtained updating the filter 4 times per second, i.e., processing chunks of m = 10000 samples of the input streams and choosing $n = 400$ filter coefficients.

C. Auscultation Management Environment

Auscultation Management Environment (AME) is a software tool constituted by a graphical user interface, shown in Fig.11, and a relational database. It allows the management, the visualization and the playback of the acquired data.

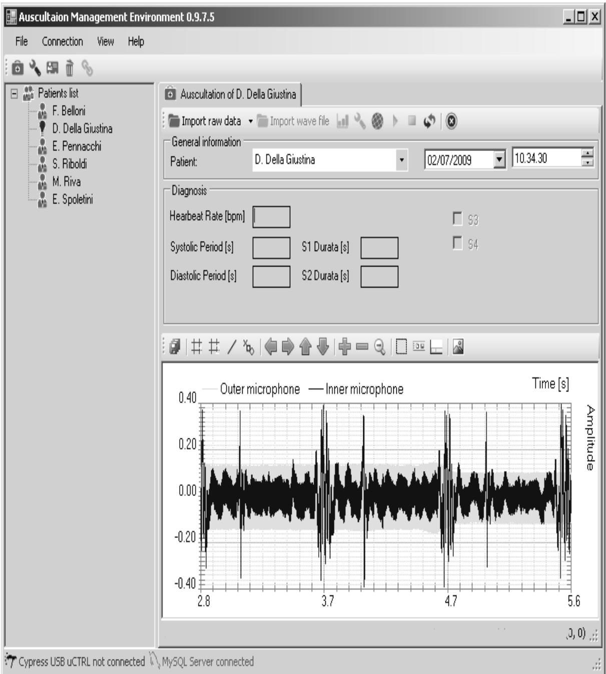

Fig.11. User friendly graphical interface of AME.

A parallel activity of the authors of this paper is focused on designing algorithms to support the automatic diagnosis of cardio-pathologies. The development of these software procedures benefits from high sound quality provided by enhanced stethoscopes. Some functionalities concerning basic cardiac measurements were already integrated in AME. Present research activities, still in progress, are focused on the development of an inferential engine that embeds the physician's ability to execute a diagnosis by hearing, as already discussed in [14] for respiratory sounds. This inferential engine requires features extracted from the cardiac sound by specific functions. Among them, the cardiac cycle isolation (end-pointing) [15]-[17] is the most important and challenging. Most of efforts are now focused to setup such capability.

III. Experimental Results

Table I and II list the main electronic devices employed in the prototypes of the AB and the PB. The demonstrator was tested in various conditions and with different types of noise, to verify the robustness of the filtering algorithm. Besides the typical environmental background noise, such as voices chatting, car motors, horns, fire and ambulance sirens, etc., synthetic noises were considered too. In each situation, a significant noise reduction ratio (NRR) is reached. NRR is defined as the ratio of the power of the signal before and after filtering. Some results are listed in Table III, where NRR is computed where the heart beat is not present.

Table I. Main electrical devices in the AB.

Transducers	Knowles Acoustic, EK-3024
Op-Amp	TI, OPA2227
PGA	TI, THS7002
Digital Isolators	Analog Devices, ADuM240x
Low Drop Voltage Regulators	TI, uA78m05,uA79m05
Low Drop Voltage Reference	TI, REF3140
ADC	TI, ADS1252

Table II. Main electrical devices in the PB.

FPGA	Xilinx, Spartan 3 XC3S200
DSP	TI, TMS320C6726B
Display	OSRAM, OS128064PK16MYO
DAC	TI, PCM1774
DSP Flash Memory	Spansion, S29AL004D
FPGA PROM	Xilinx, XCF02S
Switching Voltage Regulators	TI, TPS61202, TPS62410, TPS65130, TPS61081
Low Drop Voltage Regulators	TI, TPS72301, LP2985
Battery	Ultralife, UBBL07

Table III. Application of the filtering algorithm: some results.

Environment	**NRR dB**
Quite office: doors and windows closed; no voices	-12.95
Noisy office: doors and windows open; background voices	- 14.04
Noisy office and synthetic chirp [100 Hz – 2 kHz]	-15.50
Noisy office and low frequency noise	-13.50
Noisy office and high frequency noise	- 16.11
Noisy office and white noise	-14.48

Fig.12 shows the sounds acquired by the inner and the outer microphones when the auscultation is performed with a high frequency synthetic signal superimposed to the typical environmental noise. In such a condition, the acquired cardiac sound is unintelligible. Fig.13 shows the filtered signal after noise cancellation.

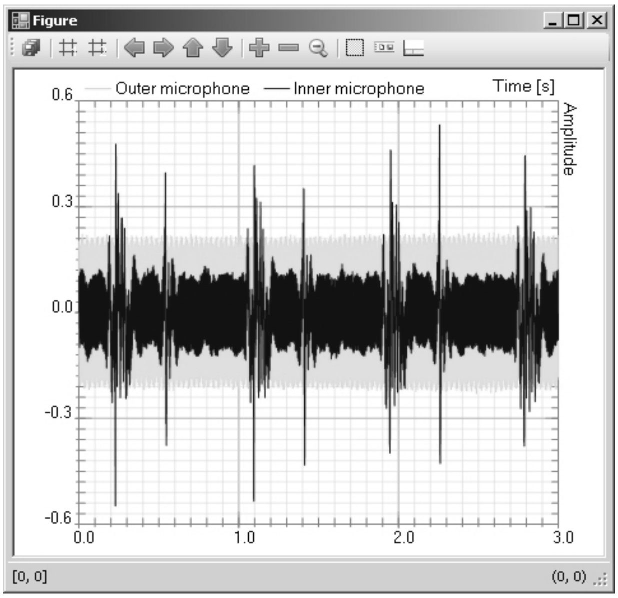

Fig.12 Comparison between signals from inner and outer microphones.

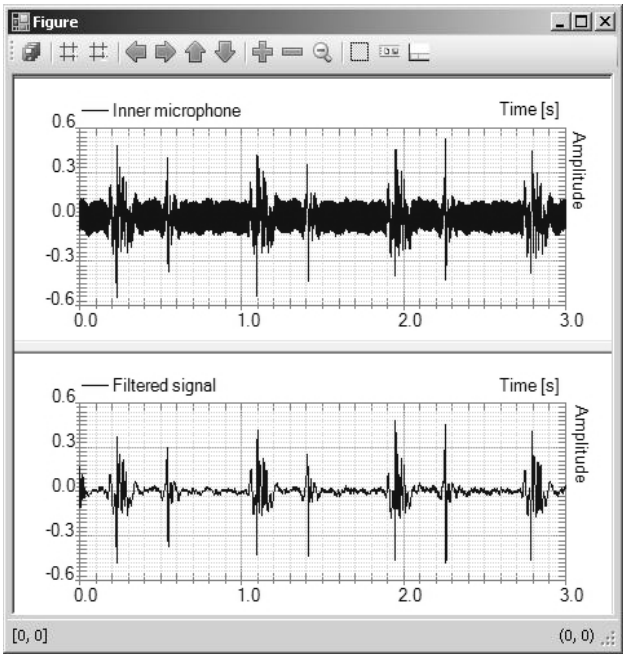

Fig.13 Comparison between inner microphone and filtered signals.

Fig.14 shows a detail of Fig.13. This technique reconstructs the sharp beat peaks while cleaning the superimposed noise at nearly the same frequency.

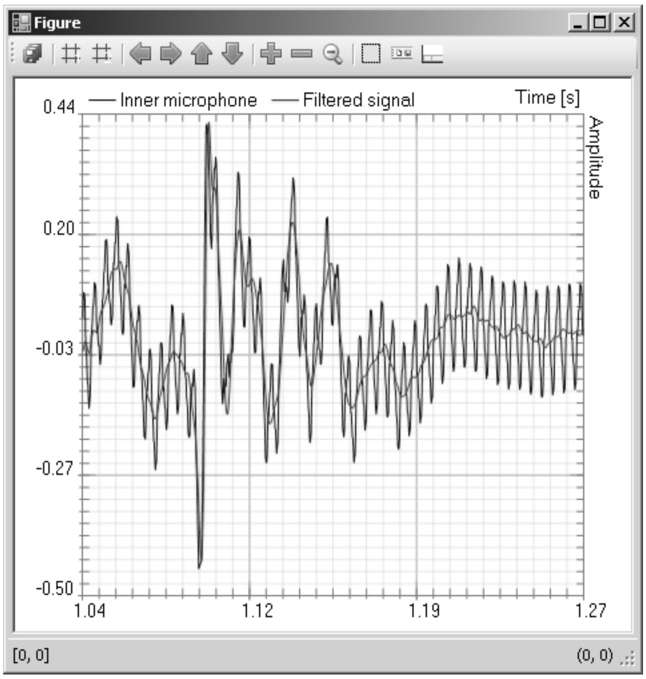

Fig.14 Detail of Fig.13.

IV. Conclusion

A new medical equipment for heart sound detection and analysis has been presented. The tool is based upon an electronic stethoscope specifically designed to improve the quality of the auscultation. Noise reduction is achieved acquiring signals from two microphones and applying a weighted difference adaptive algorithm, so that signal is not distort. The electronic hardware is enriched by a software platform to manage auscultations and to provide an assisted diagnosis. Experimental results confirm the accuracy both of the sound detection and of the noise reduction.

Appendix

Discrete-time signals can be modeled by vectors in the Hilbert space $L_2(\mathrm{Z})$ of the square integrable bilateral sequences:

$$L_2(\mathrm{Z}) := \left\{ x : \mathrm{Z} \to \mathrm{R} : \sum_{n\in\mathbf{Z}} |x(n)|^2 < \infty \right\}$$

In this set, the following functions are well defined:

Inner product: $\langle x \mid y \rangle := \sum_{\kappa\in\mathbf{Z}} x(\kappa)y(\kappa)$

Convolution product: $(x \times y)(k) := \sum_{\kappa\in\mathbf{Z}} x(\kappa)y(k-\kappa)$

Correlation product: $(x * y)(k) := \sum_{\kappa\in\mathbf{Z}} x(\kappa)y(k+\kappa)$

The energy of $x \in L_2(\mathrm{Z})$ is defined as:

$$\mathrm{E}(x) := \|x\|^2 := \langle x \mid x \rangle$$

The proposed identification criterion states that the best evaluation of the impulse response of the system is the vector $\tilde{h}_f \in L_2(\mathrm{Z})$ which minimizes the energy of the weighted difference $y := s_i - s_o \times h_f$:

$$\mathrm{E}(y; h_f) := \left\| s_i - s_o \times h_f \right\|^2 \qquad (3)$$

where $s_i = (x_h + x_n) \times h_b$ and $s_o = x_n$ are respectively the stream from the inner and the outer microphone. The former is the sum of the heart sound x_h and the noise x_n, after the interaction with the system, whose impulse response is h_b, while the latter is the noise x_n.

Proposition P1: Let $x_h, x_n, h_b, h_f \in L_2(\mathrm{Z})$ such that $x_h * x_n = 0$, then the function $\mathrm{E} = \left\| (x_h + x_n) \times h_b - x_n \times h_f \right\|^2$ has a unique minimum for $h_f = h_b$.

Proof: As first step of the proof, the previous hypothesis is used.

$$\begin{aligned}
\mathrm{E} &= \left\| (x_h + x_n) \times h_b - x_n \times h_f \right\|^2 \\
&= \left\| x_h \times h_b - x_n \times (h_b - h_f) \right\|^2 \\
&= \left\| x_h \times h_b \right\|^2 + \left\| x_n \times (h_b - h_f) \right\|^2 \\
&\quad + 2\left\langle x_h \times h_b \mid x_n \times h_b \right\rangle - 2\left\langle x_h \times h_b \mid x_n \times h_f \right\rangle
\end{aligned}$$

The two inner products can be simplified observing:

- if $a, b, h, k \in L_2(\mathrm{Z})$, such that $a * b = 0$ then also $(a \times h) * (b \times k) = 0$;
- the statistical property of null-correlation implies the geometrical property of orthogonality, in fact $\left\langle a \mid b \right\rangle = (a * b)(0)$.

This consideration allows rewriting the target function as:

$$\mathrm{E}(h_f) = \left\| x_h \times h_b \right\|^2 + \left\| x_n \times (h_b - h_f) \right\|^2$$

Being E a quadratic form, if exists a stationary point, it is the only minimum. This point fulfills the condition $\dfrac{\partial \mathrm{E}}{\partial h_f} = 0$.

Hence:

$$\begin{aligned}
\frac{\partial \mathrm{E}}{\partial h_f} &= \frac{\partial}{\partial h_f}\left\{ \left\| x_h \times h_b \right\|^2 + \left\| x_n \times (h_b - h_f) \right\|^2 \right\} \\
&= \frac{\partial}{\partial h_f} \left\| x_n \times (h_b - h_f) \right\|^2 \\
&= \frac{\partial}{\partial h_f} \sum_{\kappa \in \mathbf{Z}} \left[x_n \times (h_b - h_f) \right]^2 (\kappa)
\end{aligned}$$

$$\begin{aligned}
\frac{\partial \mathrm{E}}{\partial h_f} &= \sum_{\kappa \in \mathbf{Z}} \frac{\partial}{\partial h_f} \left[x_n \times (h_b - h_f) \right]^2 (\kappa) \\
&= \sum_{\kappa \in \mathbf{Z}} 2\left[x_n \times (h_b - h_f) \right](\kappa)(-x_n)(\kappa) \\
&= -2\left\langle x_n \mid x_n \times (h_b - h_f) \right\rangle = 0
\end{aligned}$$

Note that y because this set is closed respect to convolution product and linear combination.

Excluding trivial solution $x_n = 0$, which means absence of noise, the stationary condition implies $h_f = h_b$.

QED

This means that the more accurate is the estimation h_f of the impulse response h_b, the more efficient is the noise reduction.

Corollary: In the hypothesis of *P1* the minimum value of $s = s_i - s_o \times h_f$ is $s = x_h \times h_b$.

This function represents exactly the cardiac sound given by the acoustic stethoscopes, cleaned from the environmental noise. So, physicians do not need training different from usual to interpret auscultations.

The actual numerical treatment of the problem is instead carried out in a finite-dimension Hilbert space.

Proposition P2: Let $s_i, s_o \in \mathrm{R}^m$ and $h_f \in \mathrm{R}^n$. Then the function

$$\mathrm{E} = \sum_{k=1}^{m} \left[s_i(k) - \sum_j s_o(k-j) h_f(j) \right]^2 + \sum_{k=1}^{n} a(k) h_f^2(k)$$

is optimized by the vector h_f, solution of the linear system $Ah_f = B$, where

- $A_{ij} := \sum_{k=1}^{m} s_o(k-j) s_o(k-i) + a(i)\delta(j-i)$
- $B_{1i} = \sum_{k=1}^{m} s_i(k) s_o(k-i)$

Proof: The application of the stationary condition $\forall i \in [1,n], \dfrac{\partial \mathrm{E}}{\partial h_f(i)} = 0$ leads to:

$$\begin{aligned}
\frac{\partial \mathrm{E}}{\partial h_f(i)} &= \sum_{k=1}^{m} \left[s_i(k) - \sum_j s_o(k-j) h_f(j) \right] s_o(k-i) \\
&\quad + a(i) h_f(i) \\
&= \sum_j h_f(j) \sum_{k=1}^{m} s_o(k-j) s_o(k-i) + a(i)\delta(j-i) \\
&\quad - \sum_{k=1}^{m} s_i(k) s_o(k-i) \\
&= 0
\end{aligned}$$

QED

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