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A Novel Automatic Method to Determine Blood Pressure Based on Thresholds of Audibility

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Authors' contributions

This work was carried out in collaboration between all authors. Authors GS and MT designed the study and took the leadership on this research work. Authors GA, MM, M. Mazzotta and FM realized the electronic circuitry, performed the measurements and the statistical analysis. All authors wrote the protocol, managed literature searches, contributed to the discussion, to the test improvements, to the analysis of results and approved the final manuscript.

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ABSTRACT

We present a novel system which intends to avoid a current limit presented by the commonly adopted method to measure the diastolic and systolic blood pressure. In particular, the system allows acquiring the measurements without bothering the patient, as is with the traditional inflatable cuff necessary to restrict blood flow. In such a way, it is possible to realize a holter system that is free from the inconvenience to wake up the patient during nighttime. With an ensemble of one simple pressure sensor, one audio-frequency microphone (adopted as recommended in literature references) and a low-cost ad-hoc designed electronic circuitry, we derived data from experiments involving 20 patient records, realized in two different days. Data allowed recognizing a Gaussian relationship between pressure and audio signals. This relationship enables taking advantage from the Korotkoff sound to determine the diastolic and systolic blood pressure values.

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1. INTRODUCTION

The success of the aneroid and digital sphygmomanometer was consolidated by the overall replacement of the old devices based on mercury that was prohibited by a decree on 3 April 2009. Adhering to European community provisions, the decree aimed to reduce the risk of environmental contamination by mercury, which could have detrimental effects on both the environment and human health. Currently, the market offers different mercury-free devices for measuring blood pressure, such as aneroid sphygmomanometers, hybrid, semi-automatic and automatic ones [1]. Digital sphygmomanometers, to which this article refers to, are easy to manage for everyone, especially for elderly or heart disease patients.

In general, these devices consist of an automatic inflation mechanism, an electronic circuitry, a screen, and a storage memory. They base on the oscillometric technology, and the systolic and diastolic pressures are calculated by means of empirical algorithms. In spite of their easy usage, sometimes the digital sphygmomanometers can be of some discomfort for a number of patients or can be inconvenient when a pressure holter monitor is necessary for an entire day. This is because of the inflation mechanism, which can be poorly tolerated by someone or which causes awakening during nighttime, so that the pressure of a deeply asleep patient cannot be measured. In order to overcome these problems, our aim was to find a possible correlation, for all individuals, between the arterial pressure (measured by a pressure sensor) and the Korotkoff sounds (measured by a microphone) [2,3,4,5], so to obtain pressure values just with a low-cost microphone. Within this frame, it is not necessary the currently adopted mechanism to periodically inflates and deflate the cuff, but it can be sufficient a bracelet maintained at a previously fixed pressure. The idea of utilizing a microphone for healthcare monitoring has been then applied to the measure of the subject's heartbeat, respiration, body movement and snoring [6] but, as far as we know, this is the first application to blood pressure determination. In particular, we could predict the systolic and the diastolic pressure without traditional auscultation and technology based on oscillometry [7], but rather by extracting parameters from a reduced number of measurements of static pressure, for a low cost solution with an immediate and objective response.

2. MATERIALS AND METHODS

The idea was to replace the standard but subjective and qualitative auscultation method with a novel objective and quantitative approach. To this aim, we start transducing the pressure exerted by the arm-encircling cuff and the heartbeat sound of the patient into two distinct electrical signals. Afterwards, we determine the relationship between the two signals by means of a fitting curve [8], so to obtain information of the pressure directly from heartbeat sound signals.

Pressure and Korotkoff signals were measured by means of two different ad-hoc designed electronic circuits (Fig. 1). For the amplification and the coupling sections we adopted a dual supply (-8/+8V) Op-Amp (LM358 by Texas Instruments, Dallas, Texas, USA) with a feedback resistor $R_g=330\Omega$. The pressure was measured by means of a piezoresistive sensor (FSR by Interlink Electronics, Camarillo, CA 93012, USA) able to reduce its resistance proportionally to the applied force. Having the sensor a fixed surface, the measured force is proportional to the pressure of the cuff. According to experimental calibration, the adopted sensor was demonstrated to be capable of a sensibility as low as 4 mmHg with a linear behavior within the 0-240 mmHg range. The FSR sensor was minus 2V biased and its corresponding resistance values varied within the 120–1MΩ range. Moreover, the response of the circuit was linear within the 0-240 mmHg range of pressure.

The acoustic waves were measured by a high linearity (within the 20-18000Hz range) low-cost microphone (KEEG1538WB-100LB by Kingstate Electronics Corp, New Taipei City, Taiwan). The microphone was applied just under the cuff, following the recommendations in [9] for the microphone, that is the nearer the sound detector to the cuff the higher the sound intensity, and in [10] for the cuff, that is to place the midline of the bladder of the cuff over the arterial pulsation over the patient's bare upper arm. Coupling the microphone with a resistor, we realized a divider configuration to obtain a voltage signal. The unidirectional nature of the microphone and the difficulty of measuring the Korotkoff sounds made necessary the introduction of a mechanical amplifier realized by exploiting the terminal part of a stethoscope. The output signal was in the order of hundreds of millivolts, and therefore we accomplished an additional electronic amplification.

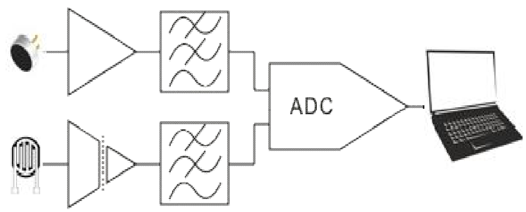

Fig. 1. Data acquisition system includes one pressure sensor and one microphone, voltage amplifiers with galvanic isolation, low-pass filters, an analog-to-digital converter (ADC) interfaced with a personal computer or any other processing device

A linear optical coupler (IL300 by Vishay Electronic GmbH, Selb, Germany) was adopted to obtain a galvanic isolation and two ICs (TL082 by STMicroelectronics, Geneva, Switzerland) were used to realize a current-controlled-current-source (V-I converter) upstream and a voltage-controlled-current-source (I-V converter) upstream of the optocoupler. Fig. 2 shows the complete circuital scheme. The resistances R1(R3) and R2(R4) determined the gain of the isolation amplifier that must be greater than one, in the bottom part of the electronic scheme in Fig. 2, to perform a correct measure of the Korotkoff sounds. The chosen values of the electrical components assured a linear behavior of the entire circuit configuration.

Data were acquired at a rate of 115200bps by means of an "Arduino Uno R3" electronic board (by Arduino, Ivrea, Italy) directly interfaced with the PC, and offline processed by homemade MATLAB® routines (Matlab® R2013b by Math Works, Natick, Massachusetts, USA). The acquisition electronic board accepted analog input signals, within the 0-5V range, from the microphone and from the pressure sensor. Furthermore, the board was equipped with a 10 bit ADC to quantize the voltage range into 0-1023 values, with a resolution of 5/1024 = 4.88 mV.

The measurements were realized in two different sessions and in two different days. Twenty informed healthy individuals performed five times each day the same test, which consisted in mimicking the auscultation method, revealing and memorizing the voltage signals coming from the measuring electronic circuit, summing 200 observations in total.

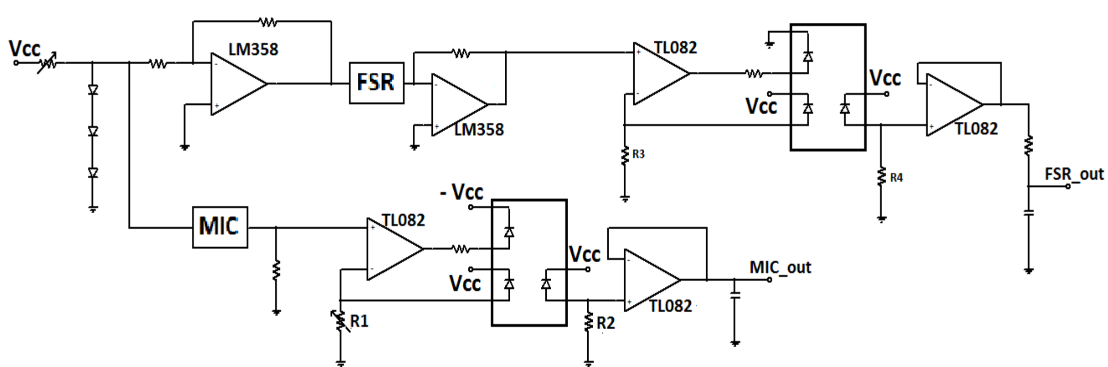

Fig. 2. Electronic conditioning circuitry to record signals from the pressure sensor and microphone. A voltage regulator implemented by a three-diode structure polarized by a trimmer resistance. The voltage regulator drives the microphone and the pressure sensor with an inverting buffer. Two optocouplers make the circuit galvanic isolation with an input CCVS and an output VCCS. A second trimmer resistance guarantees the correct function of microphone in the linear region of the optocoupler. Both output signals are low-pass filtered and sent to the computer by a serial connection

At once, measurements were recorded by the sphygmomanometer and the stethoscope and, as a reference, an operator revealed the blood pressure by auscultation. In such a way, we were able to compare the responses from the microphone and the pressure sensor with the systolic and the diastolic pressure, so to determine a relationship between the methods of measurement, if any.

3. RESULTS AND DISCUSSION

Data analysis allowed recognizing the threshold of hearing that was transduced into a voltage signal by making a comparison with the diastolic and the systolic pressure. A first adopted method, said *first strategy*, was the statistical evaluation of this voltage value to estimate its repeatability for each measurement, as shown in Fig. 3.

Averaging two hundred measures, we obtained a value for the threshold of audibility equal to 209, with a standard deviation of 20.52 (34.6%). Based on these results, we hypothesized the threshold of hearing to be related to the amplitude envelope of the audio signal. Being the envelope different among the individuals because of the different measurement conditions (i.e. different persons, different time, different stress, and so on), it was chosen a second method, said *second strategy*, to normalize the voltage value, which represents the threshold of hearing, to the maximum peak of the waveform peculiar of each person. We statistically evaluated the voltage related to the threshold of hearing normalized to the maximum amplitude of each set of data and, after making again an average of the two hundred measurements, we obtained a normalized average value for the threshold of audibility that was equal to 1.2059, with a standard deviation of 0.0888 (7.36%). We believe the *second strategy* to be more effective being each measurement strictly related to the physiological condition of the patient.

The achieved results were used to estimate the diastolic and systolic pressure of each individual and to verify the error in measurement being within 0-4 mmHg range. This estimation was performed with a graphical method. Three lines were drawn, one for each measure, parallel the x-axis representing the pressure.

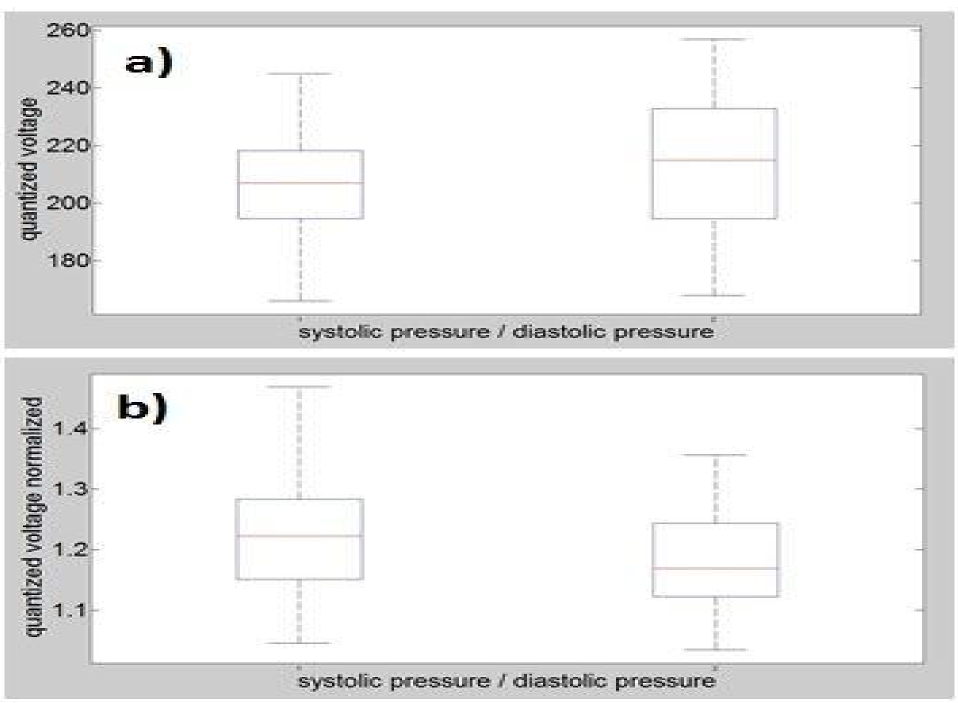

Fig. 3. The median value of the voltage relative to the systolic (on the left side) and to the diastolic pressure (on the right side), respectively obtained through the (a) first and (b) the second strategies

The corresponding intercepts at y-axis were, on the basis of the first strategy $y_1=m_{th}=209$; $y_2=m_{th}+\sigma=209+20.52$; $y_3=m_{th}-\sigma=209-20.52$ and on the basis of the second strategy $y_1=m_{th-n}=1.2059$; $y_2=m_{th-n}+\sigma_n=1.2059+0.0888$; $y_3=m_{th-n}-\sigma_n=1.2059-0.0888$, where:

- m_{th} represents average voltage value related to the threshold of hearing;
- m_{th-n} represents average voltage value normalized related to the threshold of hearing;
- σ represents the standard deviation related to m_{th};
- σ_n represents the standard deviation related to m_{th-n};

Fig. 4 shows the graphical method applied to two measurements as an example. The points of intersection between the straight line, which specify the average voltage value related to the threshold of hearing, and the amplitude envelope of the audio signal, identify the systolic and the diastolic pressure. The measurement error was estimated by means of the intersection points between the other two straight lines and the amplitude envelope of the audio signal.

In the first example with the *first strategy*, the diastolic pressure was determined to be 85±6 mmHg and the systolic pressure was 114±5 mmHg, while with the *second strategy*, the diastolic pressure was 80±2 mmHg and the systolic pressure was 115±3 mmHg. In the second example the *first strategy* furnished the diastolic and systolic pressure of 65±6 mmHg and 106±6 mmHg respectively, while the *second strategy* gave results as 68±2 mmHg and 105±4 mmHg respectively.

The graphical method applied on the entire data allowed to obtain the following averaged measurement errors: with the *first strategy* 12 mmHg and 9 mmHg for systolic pressure and diastolic pressure respectively; with the *second strategy* 4 mmHg and 2 mmHg for systolic pressure diastolic pressure respectively.

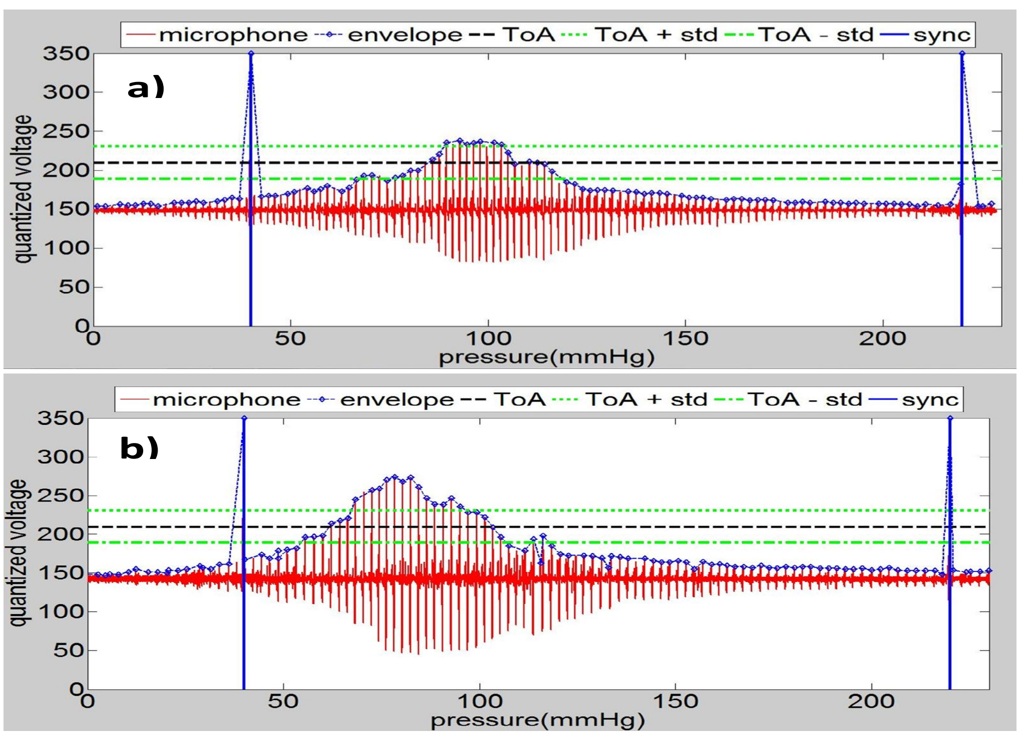

Fig. 4. Examples related to a single measure (two vertical spikes referencing known pressures of 40 mmHg and 220 mmHg respectively, for synchronization purposes). DP= diastolic pressure, SP= systolic pressure. (a) First strategy: DP=85±6 mmHg, SP= 114±5 mmHg; Second strategy: DP=80±2 mmHg, SP=115±3 mmHg. (b) First strategy: DP=65±6 mmHg, SP=06±6 mmHg; Second strategy: DP=68±2 mmHg, SP=105±4 mmHg

We then considered the mathematical approximant of the amplitude envelope of the acoustic signal as a function of the pressure, so to take out the systolic and the diastolic blood pressure without performing further measurement. The best result was obtained with a Gaussian function:

$$M * e^{-\frac{(x-m)^2}{2\sigma^2}}$$

where:

- M is the maximum peak of the amplitude envelope of the audio signal;
- m is the pressure value that corresponds to the maximum peak of the amplitude envelope of the audio signal;
- σ is the standard deviation and relates to distance between the diastolic and systolic pressure.

Fig. 5 shows two examples of approximate vs. measured curves related to two different individuals. The errors made on diastolic and systolic blood pressure values were of 0.8 mmHg and 3.3 mmHg for the first, and 2.06 mmHg and 2.6 mmHg for the second individual respectively.

Aforementioned data regard single examples in a group of 20 volunteers under test. For completeness purposes, Table 1 reports all the averaged ± standard deviation values of diastolic and systolic pressures, with comparison results for first and second strategies respectively. The system demonstrated reliable results.

Based on the obtained results, the Gaussian model for the sound signal acquired from a microphone inside a sphygmomanometer furnished better results for the diastolic pressure with respect to the systolic one, its accuracy depending on the physiological condition of the patient. Besides, the audio signal presented asymmetry; in fact, the waveform was better detectable at low pressures rather than high pressures, presumably because an arm at high pressures moves towards ischemia conditions. As a result, the approximation curve increases its interpolation error toward higher pressures but remaining quite close to the audio waveform anyway.

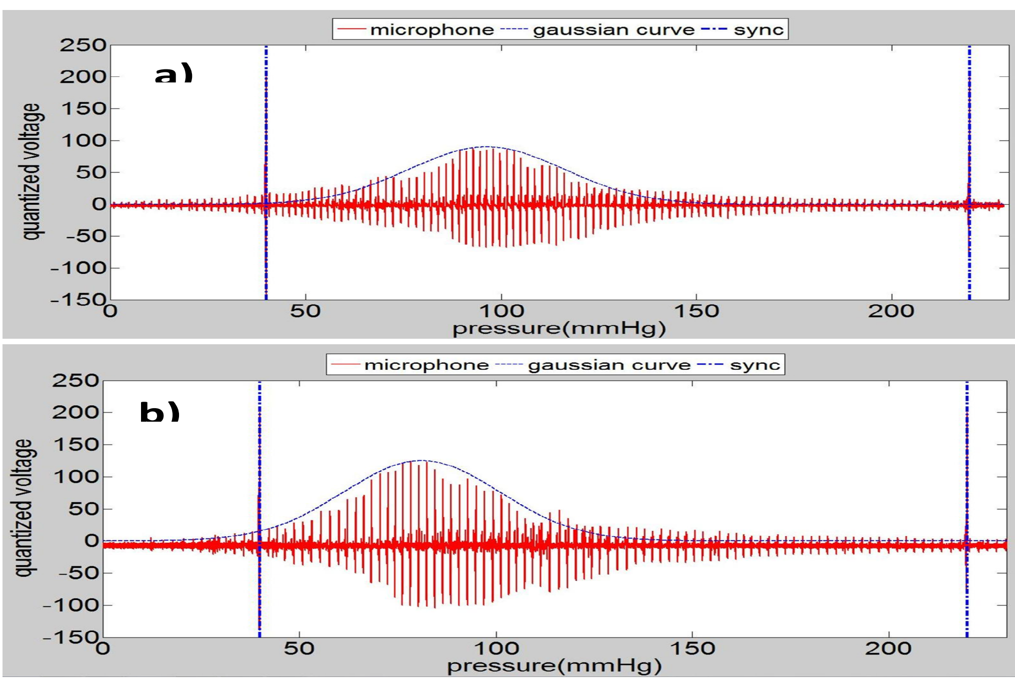

Fig. 5. The audio signal related to blood pressure and the slow varying line representing the interpolating Gaussian curve. The graphs are related to two different individuals. The two spikes ('sync' in the legend) reference a known pressure. (a) The Gaussian curve approximates the amplitude envelope and detects the diastolic blood pressure with an error of 0.8 mmHg while the systolic pressure with an error of 3.3 mmHg. (b) The error related to the diastolic pressure was 2.06 mmHg, the error related to the systolic pressure was 2.6 mmHg

Table 1. (DP: Diastolic Pressure, SP: Systolic Pressure): Averaged ± standard deviation values obtained by means of the first and the second strategy for all subjects

Subject	First strategy		Second strategy	
	DP [mmHg]	SP [mmHg]	DP [mmHg]	SP [mmHg]
#1	67±6	106±5	65±3	105±2
#2	55±10	107±8	50±3	120±2
#3	73±6	110±5	70±3	112±2
#4	70±4	111±6	72±2	110±4
#5	82±5	114±5	78±3	116±3
#6	65±5	108±4	63±2	110±2
#7	60±6	102±5	62±2	105±3
#8	53±5	105±5	55±2	103±4
#9	51±6	100±5	54±3	101±4
#10	78±4	122±6	75±2	120±3
#11	57±3	109±6	61±4	108±3
#12	72±6	112±7	73±2	110±3
#13	78±7	121±7	75±2	119±4
#14	71±6	118±5	69±3	120±3
#15	69±5	116±5	71±2	115±2
#16	65±4	105±6	65±4	106±2
#17	70±8	113±5	71±3	115±4
#18	77±6	115±5	75±2	120±3
#19	50±7	125±10	47±3	125±4
#20	60±5	108±4	62±3	112±2

4. CONCLUSION

We designed an electronic conditioning circuitry to acquire data from both a pressure cuff and a low-cost audio microphone. From data comparison we were able to justify that a reliable blood pressure measure can be obtained from acoustic signals alone.

In particular, we used two different strategies to measure the diastolic and systolic pressure based on the estimated voltage value related to the audibility threshold. The *first strategy* based exclusively on the voltage value, whereas the *second strategy* adopted the voltage value but normalized to the maximum peak of the amplitude envelope of the audio signal related to each individual. In both cases, it was possible to predict the diastolic and systolic pressure for all individuals under test, but the *second strategy* lowered the estimation error.

The number of the individual performing the test, the timing session and the measures taken in each session were enough to guarantee a variety of situations able to analyze both the device and the methodology, considering the present work as a preliminary stage of the research. Mimicking the same procedure used by human operator, we obtained measures with significant results, that it is possible to obtain a useful resolution in blood pressure measurement. In fact measures taken with a discussed methodology resulted with an error below 4 mmHg, which is a value historically considered reasonable low. Furthermore, it resulted possible to interpolate measured points with a Gaussian curve, which was not an obvious result at all. With respect to our proposal, the common adopted human auscultation method holds differences in any case, both because it is subject-related and because of the non-linearity of the ear response. We found how a Gaussian curve result as the best fit for every experimentally acquired measure, with an approximation error always remaining below the measure error. Within this frame, an interpolation curve can be considered instead of the actual data. We found that the sets of data presented few error points caused by mechanical disturbance of the microphone (rubbing with tissue for example), but the measure noise was negligible in any case. The repeatability of the measures was obtained within different subjects too.

The basic advantage of the proposed system is to furnish the possibility to perform measurements without bothering a patient so to realize a holter device without the disadvantage to wake up the patient during nighttime. In addition, this system is designed to have the ability to continuously measure the diastolic and systolic pressure, without applying the auscultation method, by means of static pressure measurements. It is possible to avoid recurring to periodically inflate and deflate a dedicated cuff, but it is possible to simply recur to a bracelet, tight to impose a static known pressure, so that through a microphone is possible to detect the peak amplitude at this pressure. Interpolating the peak points obtained and applying the implemented algorithm, we are able to calculate systolic and diastolic pressure of the patients.

The results were obtained on twenty peer students as testers, but incoming works will investigate other population such as athletes, children, and elderly or heart patients.

ETHICAL APPROVAL

All authors hereby declare that all experiments have been examined and approved by the appropriate ethics committee and have therefore been performed in accordance with the ethical standards laid down in the 1964 declaration of Helsinki.

COMPETING INTERESTS

Authors have declared that no competing interests exist.

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