Continuous Noninvasive Arterial Blood Pressure Monitor with Active Sensor Architecture

Viacheslav Antsiperov
Department of Physical and Quantum Electronics
Moscow Institute of Physics and Technology
Dolgoprudny, Moscow Region, Russia
e-mail: antciperov@cplire.ru

Gennady Mansurov
Laboratory of Diagnostic Systems
Kotel'nikov Institute of REE of RAS
Moscow, Russia
e-mail: gmansurov@cplire.ru

***Abstract*—The paper presents the latest results of developing a new method of noninvasive continuous blood pressure monitoring. This method is based on the principle of pulse wave compensation. It is shown that sensors for such a measurement should be not only smart, but also active. In this connection, the concept of smart sensors is expanded to the concept of active sensors. The technical design of the active sensor for noninvasive pressure measurement is described. The results of active sensor calibration and testing are under discussion. The last section of the report is devoted to the development of software for active sensor control - its intellectual stuffing. In this section is described and justified a new principle of active measurement of quasi-periodic processes - pulse wave compensation based on prediction patterns. The progress achieved in the research and the ways for further investigation are outlined in the conclusion.**

Keywords - smart and active sensors; compensation method; arterial blood pressure monitoring.

I. INTRODUCTION

Advancements in semiconductor industry substantiate a dramatic growth of rate of innovations in a wide range of human activities - from computer technology and communications to home appliances and transport. This acceleration is due mainly to delivering smaller, cheaper, more reliable solutions able to integrate critical functions such as analog circuits, and embedded microprocessors in a single device. New packaging technologies and methodologies have also improved the cost and reliability of the assembly of packages with small footprints. For example, chip-scale packages have reduced semiconductor package sizes to the dimensions of the die inside. Other innovations, such as the use of a known good die or foldable printed circuit boards, have also provided considerable miniaturization in the overall space occupied by the electronic portion of the modern devices.

Medical technology is a broad field where the above innovations play a crucial role in sustaining health. The development of medical devices and equipment has made significant contributions to improving the health of people all around the world. From "small" innovations like healthcare gadgets that help a person to monitor and manage serious health conditions like asthma, heart problems (see Fig. 1), diabetes, etc., to larger, more complex technologies like MRI machines, artificial organs, and robotic prosthetic limbs, technology has undoubtedly made an incredible impact on medicine.

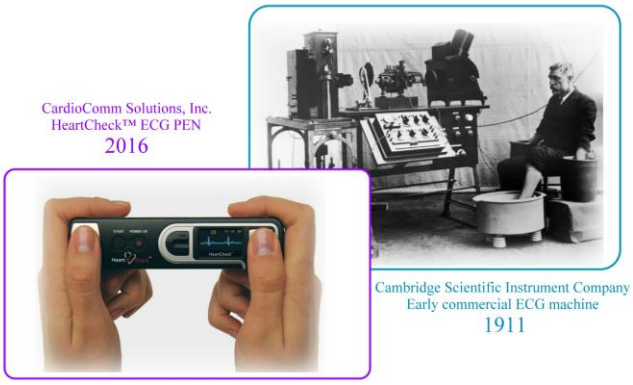

Figure 1. An example illustrating the incredible progress in medical technology: the ECG monitor evolution from the first commercial huge size machine (1911), consuming a lot of electrical energy, to a pocket-sized healthcare gadget (2016), powered by a compact battery.

It should be noted that the development of low-cost, low power, multi-functional medical devices, that are small and can communicate in wireless manner over short distances, like ECG monitor shown in Fig. 1, is only the visible tip of the modern medical technology iceberg. Its main core is that medical devices are getting smarter. Embedded microprocessors, SOC (System On a Chip), SBC (Single Board Computers), integral PDAs (Personal digital assistants) and advanced sensor technology are all common elements of today's medical devices. The computing power of microprocessors and high performance peripherals provide the ability to support several simultaneous processes: the sensor management, signal processing, data collection and analysis, wired and wireless communications and user interface. This gives unlimited possibilities in the design of medical devices. On the other hand, it leads to myriad ways of combining the components in developing well-defined purpose devices, and, accordingly, it requires to investigate a lot of possible solutions and to find the optimal one.

In this paper our experience in the development of a modern high-tech medical device - arterial blood pressure

(ABP) monitor is presented and the results obtained in its implementation, including the main result - the elaboration of the basic architectural components concept, named an Active Sensor by analogy with the Smart Sensor (see Fig. 2) are discussed. Initially, these results have been briefly considered at the Second International Conference on Smart Portable, Wearable, Implantable and Disability-oriented Devices and Systems (SPWID 2016) [1]. In this article they will be discussed in detail.

The structure of the paper is as follows. Section II provides a detailed discussion of the Smart Sensor and the Active Sensor concepts. It also gives strong arguments in favor of the Active Sensor concept for developing medical devices. Section III gives a short overview of existing non-invasive ABP monitoring methods. Section IV introduces a new method of ABP monitoring proposed by the authors – *pressure compensation method.* Section V describes in detail the structure and design of the sensor developed. In Section VI the results obtained in real experiments with active ABP monitoring are reported. In Section VII some ways to overcome the shortcomings associated with the incomplete compensation of time–varying blood pressure caused by a simplified structure of the PID (proportional–integral–derivative) feedback controller are scheduled. Finally, the Conclusion briefly summarizes the problems discussed in the article.

II. FROM SMART SENSORS TO ACTIVE SENSORS

One of the main trends in developing modern devices and instruments in medicine as well as in a rather broader field – the development of industrial control and monitoring systems – is a full implementation of smart sensors [2].

The basic architectural components of the Smart Sensor are shown in Fig. 2 (a more detailed description can be found in IEEE 1451 Standards, created by the IEEE Instrumentation and Measurement Society's TC-9 Technical Committee on Sensor Technology). The Smart Sensor consists of four main components: the *Sensing Unit* (SU) that collects real-world analog data, the *Signal Conditioning* circuitry and the *Analog to Digital Converter* (ADC) destined to condition and convert data to the digital domain, and, finally, *microprocessor* (μP), which by means of Applications installed performs the main tasks of the Sensor. Task processing can be supported by *Memory* and *Communication Unit*; some results can be displayed by *Local User Interface* (IEEE 1451 general model of a Smart Sensor).

The most important component of the Smart Sensor architecture is certainly a microprocessor (μP). As mentioned above, the computing power of microprocessors provides both the ability to support multiple simultaneous processes and the processing of data associated with each of them. Data processing can be very complicated: nonlinear, adaptive to data variability, incorporating artificial intelligence (AI) elements, etc. However, it should be noted that due to the unidirectional data flow from SU to the microprocessor, the latter does not affect the conditions of data measurement – it can only manipulate in some way the data measured.

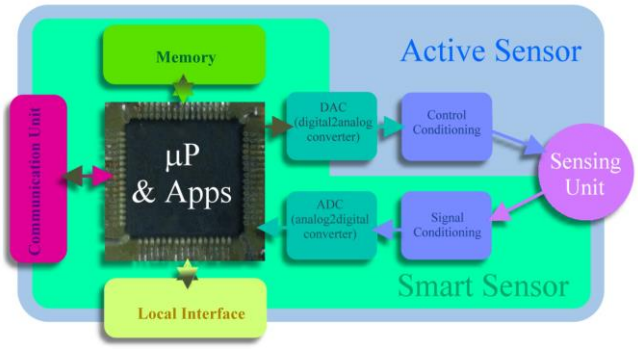

Figure 2. The principal components architectures of Smart Sensor (green) [2] and Active Sensor (blue) [1]. The latter extends the circuit of Smart Sensor by adding the Sensing Unit control sub-circuit.

The above fact is of little importance in artificial systems, whose behavior is stationary or is forecasted at least statistically. The situation is different when somebody deals with living systems, e.g., with a human. As noted in [3], due to highly non-linear, non-equilibrium nature of living systems even the slightest differences in the physiological state of the bio-system used and in the conditions obtained in a experiment can, in consequence of deterministic chaos, assume singular importance.

We came face to face with this situation and its inherent problems when a few years ago began to develop a smart medical device – arterial blood pressure (ABP) monitor. At that time we thought that supplying the classic tonometer by some tracking system, followed by block forming pressure dynamics prediction and flexible active sensor controller based on the prediction, we will be able to monitor ABP continuously in time in absolute units – in mmHg. It is possible because such a device would have to implement the proposed earlier *compensation method* for measuring the pressure. Of course, to compensate the variable pressure in real time that monitor had to include powerful computational tools. We believe, that the use of embedded microcontroller with smart applications can produce much better results than a standard digital tonometer can. So, the use of a built-in microcontroller (μC) and its periphery gives the monitor characteristic features of smart sensors, as shown in Fig. 2. But it turned out, that for the realization of this idea we had to go even one-step further – to introduce a SU control feedback loop, which depends on data observed. Thus, the idea of an Active Sensor was created, whose components architecture expands the architecture of the Smart Sensor.

The basic architectural components of the Active Sensor are shown in Fig. 2. The Active Sensor circuit extends the circuit of the Smart Sensor by adding the Sensing Unit control sub-circuit. In contrast to the data measurement path, the control sub-circuit is directed in the opposite way - from the microprocessor to SU, although composed essentially of the same, but performing the inverse functions components - *Digital to Analog Converter* (DAC) and *Control Conditioning* circuitry. Thus, if in the Smart Sensor SU is

simply, in unidirectional way connected to the microprocessor, in the Active Sensor it is included in a loop circuit consisting of the data measuring and controlling sub-circuits.

Before moving on to the practical implementation of the above rather abstract concept of an Active Sensor, it is appropriate to review briefly the set of ideas and approaches existing today in the field of non-invasive arterial blood pressure monitoring.

III. ABP MONITORING APPROACHES

One of the first (1876) noninvasive devices suitable for measuring human arterial blood pressure was a plethysmograph invented by Etienne Jules Marey. Its functioning was based on the recording of pulsed changes in the limb volume caused by ABP pulsations [4]. Upon measurements, the patient's forearm was placed into plethysmograph, which was filled with warm water. A change in the blood content in the forearm arteries resulted in water displacement from the plethysmograph, which in turn moved the float and stylus that recorded the plethysmogram, as shown in Fig. 3.

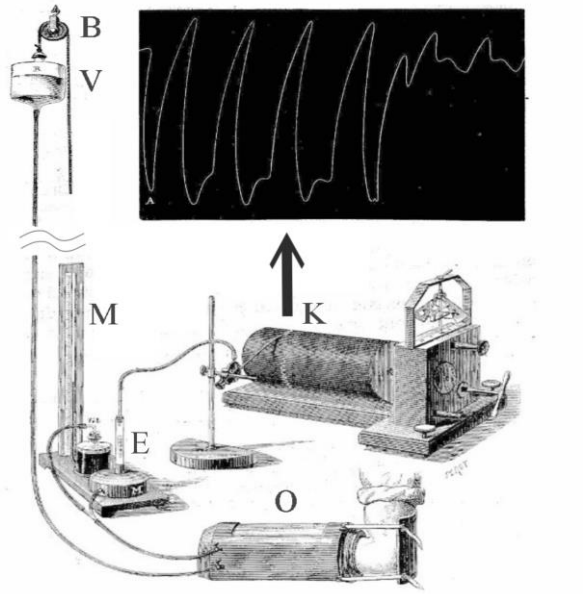

Figure 3. Marey's device for measuring ABP: the plethysmograph O is connected with both reservoir V that is filled with water and hung up on a block B and a mercurial manometer M; the volume change is written in sooty tape K by a stylus connected with float E [4].

In the Marey's experiment, volume changes were recorded at constant pressure in the plethysmograph O, which was set by the height level of water reservoir V and reflected in monometer M connected to O. In his experiments, Marey obtained many interesting results; he found the dependence of the plethysmogram amplitude on the plethysmograph pressure [4]. In principle, on the bases of these data, as it became known later, Marey could have determined the two main parameters of ABP, both systolic and diastolic pressure. However, due to difficulty of the implementation and ambiguity of interpretation of the result, Marey's plethysmographic method was not used for a long time (about 100 years). At the end of the 19th century, more attention was paid to methods of measuring ABP that are based on sphygmomanometry, monitoring of changes in the character of the heart pulses (tense pulse, soft pulse, etc.), which is manifested in micromovements of arterial walls. The main task of sphygmomanometer devices was to find such an external pressure on an artery that would balance the counterpressure of the blood with the bloodstream ceasing and the pulse, correspondingly, disappearing. The most successful sphygmomanometer out of the first devices was the device of Samuel von Bashe, created by him in 1881. With its help he determined systolic pressure, although only approximately. Gustav Gartner in 1899 improved this device and referred to it as a tonometer.

The search for a noninvasive way to measure ABP culminated in two discoveries that made successes in the cardiological studies in the 20th century possible: description of Riva-Rocci's sphygmomanometer in 1896 and discovery of Korotkoff's sounds in 1905. Riva-Rocci's method is compression of a brachial artery using a special cuff that relates to a mercurial manometer. Air is pumped into the cuff using a bulb until the pulse in a radial artery disappears and, then, it is slowly released. ABP is determined by manometer indications, when the pulse appears. Thus, the device makes it possible to assess systolic pressure. Korotkoff's discovery of the particularities of sound effects upon decompression of a brachial artery in 1905 underlies an alternative method for determining systolic and diastolic ABP. It is believed that Korotkoff's sounds relate to impairments in the laminar flow of blood in arteries upon their crossclamping. This impairment is accompanied by shock phenomena, and their sound makes it possible to hear the pulse within the range between systolic and diastolic ABP.

Technological advances in the field of microelectronics and computer engineering made it possible in the second half of the 20th century to improve Riva Rocci's and Korotkoff's classic method for measuring ABP [5]. The improvements were mainly connected with the intention to maximally exclude the effect of the human factor on measurements. Thus, it was found that the most convenient estimate of the pulse for automatic procedures of determination of parameters of ABP is the amplitude of pressure oscillations in an occlusion cuff, a fact that was discovered by Marey a hundred years before! In 1976 the first bedside automatic instrument for measuring ABP (Dinamap825), which employed a modified method of Marey's, was produced. The transition from a plethysmograph to a typical occlusion cuff combining a device for creation of external compression and a pulsation sensor, which is not very accurate, but acceptable for measuring ABP, gave birth to a new oscillometric technique. Today, a portion of automatic and semiautomatic AP meters that are based on the oscillometric technique constitute approximately 80% of all the devices. Manufacturers intensely compete in the field of study of algorithms for processing measured data to increase the accuracy and reliability of the results obtained. Recent

devices have an increased intelligence level (including fuzzy logic models), and they could be fully related to smart sensors (see Fig. 2).

The methods mentioned assume a single measurement of the ABP parameters. It is often sufficient for general assessment of the state of the cardiovascular system. However, there are many situations that require continuous beat-to-beat ABP monitoring. At the worst, it is implemented invasively directly inside one of the peripheral arteries upon introduction of a catheter into it. However, since invasive methods relate to patient discomfort and there is a risk of complications upon introduction of a catheter, noninvasive methods are preferable for routine measurements.

Jan Peňáz was among the first to put forward a method of continuous non-invasive blood pressure monitoring [6]. His method described in 1973 was aimed at reducing the risks of arterial cauterization. The Peňáz method employed the idea of "vascular unloading", based on the assumption that in the "unloaded state" the pressure inside the blood vessel is equal to the outside pressure.

The basic element of the device proposed by Peňáz is a small finger cuff (Fig. 4 (A)) that has an infra-red (IR) light source on one side and a light receiver on the opposite side. The blood volume of the finger is estimated via the absorbance of IR light. The signal obtained by such a plethysmograph is further used in a feedback loop to control the pressure in the cuff. The pressure is controlled in such a way that blood volume in the finger is kept constant in time and equal to the volume, which corresponds to the unloaded vessels state defined during the calibration process. In this case the oscillations of the controlling pressure are approximately equal to pressure in the arteries. Later some formulae were proposed for recalculating pressure from finger vessels to brachial arteries, which made it possible to verify the method with respect to classical procedures.

Figure 4. Modern approaches to ABP monitoring systems.

Another technique, which provides continuous non-invasive blood pressure monitoring is arterial tonometry [7]. Like the Peňáz method arterial tonometry is based on pulse oscillation estimates, but here the principal of arterial unloading is different. In this case the cuff is placed on the wrist, so the sensor is over the radial artery (Fig. 4 (B)). The sensor presses the artery to the radial bone until it is flattened enough but not occluded. At this intermediate position arterial wall tension becomes parallel to the tonometer sensing surface and arterial pressure is then the remaining stress (perpendicular to the surface) measured by the sensor. The pressure needed to flatten but not occlude the artery is known as the "proper hold-down pressure" and is calculated by a complicated algorithm, which includes the preliminary estimate of systolic, diastolic and pulse pressures over a range of "hold-down pressures".

Currently, the devices employing these methods include the CNAP^{TM} (Peňáz's approach, Fig. 4 (A)) and T-line from Tensys Medical (arterial tonometry approach, Fig. 4 (B)).

The methods of continuous non-invasive blood pressure monitoring have both advantages and disadvantages [5]. We believe that the main drawback of these non-invasive methods is the following: irrespective of the method of vascular unloading the control of the unloading is exercised using integral parameters (blood vessels filling, overall sensor force, sensor displacement, etc.). It enables monitoring an average pulse wave of ABP but does not guarantee the details of the pulse form. We proposed a new method of arterial blood pressure monitoring that is aimed at local unloading of arterial walls by compensating local pressure.

IV. Compensation Based Approach to ABP Monitoring

When analyzing well–known methods of non–invasive continuous measurement of blood pressure, we conclude that the best results of monitoring non–stationary dynamics of blood pressure are achieved by the so–called compensation methods or methods like them.

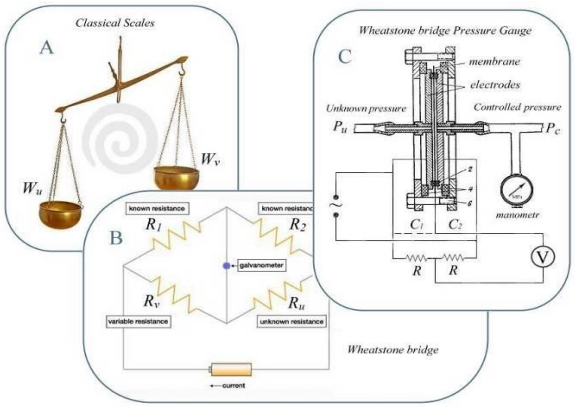

Figure 5. Measuring physical quantities using the compensation method: (A) measuring object weights W_u using a balance scale; (B)a bridge circuit of measurement of unknown resistance R_u; (C) measuring pressure P_u in aggressive media by controlling the undeformed state of a membrane.

Compensation methods are applied for measurement of various physical quantities and are based on the compensation of an unknown measured value by controlled counter value and nullification of their difference. The simplest example of the compensation method is the use of balance scales on which unknown mass W_u is measured

using a set of weights W_v, Fig. 5 (A). A scale null indicator is a vertical position of a beam or connected arrow pointer.

The compensation methods are high-precision, applied for mechanical and electrical measurements, and can be implemented in the form of bridge and half-bridge circuits (Figures. 5 (B), 5 (C)). In some cases, to increase measurement precision at deviations from zero that are less than the discreteness of a controlled physical quantity, a calibrated scale of a null detector can be used. Note that compensation methods are usually used to measure the static variables – constant unknown resistance R_u in bridge circuit – Fig. 5 (B) and the steady pressure P_u in aggressive environment – Fig. 5 (C).

We consider the compensation method as the fundamental basis to measure the varying blood pressure. The application of this method for measuring the non–static dynamic quantity has become possible for two reasons. First, the fact that blood pressure changes are not so fast, its rhythm is of the order of one beat per second, and its spectrum fits into the range of a few tens of Hz. Second, there are relatively cheap high-performance microcontrollers (ATMEL, MICROCHIP, STMicroelectronics, etc.) available at present, for which the ABP dynamics represents almost quasi-static changes.

V. The Structure and Design of the Sensor

As noted above, our method of measuring ABP is aimed at local, without cuffs, unloading of arterial walls by means of compensating intra–arterial pressure by controlled pressure. Our method of measurement is like the method of bridge measurement of unknown pressure P_u in Fig. 5 (C), but it has dynamic, "adaptive" features as in the Peňáz method [6]. The appearance of the device developed for continuous ABP monitoring and its typical use for measuring intra–arterial pressure is presented in Fig. 6.

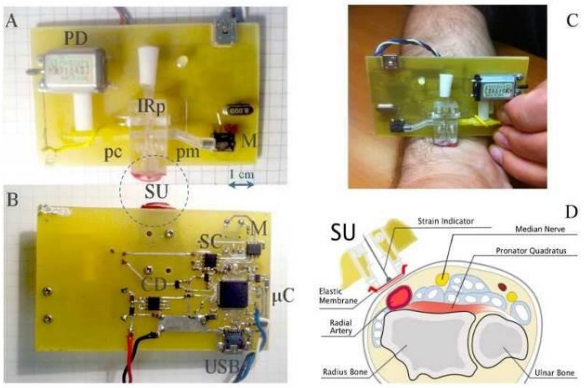

Figure 6. Active sensor for continuous blood pressure monitoring with a fluid-filled SU camera and an aperture covered by an elastic membrane; its typical arangement on the patient's wrist.

Note that the device developed (Figures 6 (A), 6 (B)) conforms exactly to the concept of an Active Sensor (see Section II, Fig. 2). The main role here is played by the programmable microcontroller µC (STM32L152RBT), that has the communication I/O (USB) on one side, and on the other side it has *pm* (pressure monitor – data measuring sub-circuit) and *pc* (pressure control – SU controlling sub-circuit) interfaces to the measuring unit SU (sensor unit).

The data measuring sub-circuit *pm* contains membrane displacement indicator SC (signal conditioning) and pressure gauge with instrument amplifier M. The SU controlling sub-circuit *pc* includes the pump PD and the chip for pump control CD (control driver). Both sub-circuits – *pm* and *pc* – are tubes coming from the fluid-filled cavity of the measuring unit SU. SU is a camera with an aperture covered by an elastic membrane (of red color in Fig. 6(D)) and containing a thin rod with one end attached to the membrane, the other end partially overlaps IR-radiation flux inside optoelectronic infrared pair IRp. Thus, this rod gives us a way to measure the membrane deformation as it serves as an indicator of displacement.

When the pressures in the SU chamber and directly behind the membrane differ, the latter is deformed with respect to one side or another and the rod is shifted in the direction of the deformation. This displacement leads to a change in the shuttering of the radiation flow IRp, which is then recorded by the displacement indicator SC. If the SU pressure is chosen so that the membrane returns into a flat, undeformed state, then the pressure out of SU can be measured due to the equality of the pressures on both sides of the membrane. If SU is established directly above the radial artery as it shown in Figures 6 (C), 6 (D), then we measure the arterial pressure under the assumption that it is equal to the external pressure. This constitutes the idea of the compensation method that we proposed.

Our last improvement of the ABP monitor presents a somewhat different technological SU implementation. Here a fluid, filling SU camera is substituted by the air. We had to make this step as the previous construction of the ABP monitor had to be supplied by sealed SU camera to prevent fluid leakage. To prevent the permeability, it requires to supply the monitor by additional control tools and additional mechanical components, which could bring about a considerable complication of device setup. Thus, it was decided to develop a monitor with a chamber filled with air, because small air leakage being acceptable in this case.

Moreover, it turns out that a substitute of working liquid by air opens new ways of SU construction implementation – a controlled air leakage could be used as a toll controlling pressure in the SU chamber. It turned out that in this case it can be done without the elastic membrane covering the aperture! The skin surface within SU aperture could act as a membrane.

The appearance of the ABP monitor with air filled SU chamber and its general structure are shown in Fig. 7. Besides the main chamber with compensating air pressure P over aperture (similar in construction to fluid filled SU chamber) there are channels diverting air with their chambers and pressure P_{out} gauges. These channels are located near the aperture on the surface of the SU contacting with wrist. Thus, here like in the previous version of ABP monitor is possible to pick out two sub-circuits – pm (data measuring sub-circuit) and pc (controlling sub-circuit). The pm sub-

circuit by means of an outer pump controls the compensating pressure P in a main chamber of SU. The pc sub-circuit measures the difference $\Delta P = P - P_{out}$, based on which the control is carried out. The control is aimed at providing such meanings of ΔP, that ensure minimum air flow from aperture to the channels diverting air. This implies the existence of the thinnest channel between the SU surface and the surface of wrist. In other words, the aim of the control is to keep the dynamics of the compensating pressure in such a way that the surface of the skin would be separated from the SU, but would not move away from it, as shown in Fig. 7 (C). In general, construction described reminds a sensor for classical tonometry forward in [8], but the advantage of our ABP monitor is that it enables to make direct pressure measurement and it does not have mobile mechanical parts.

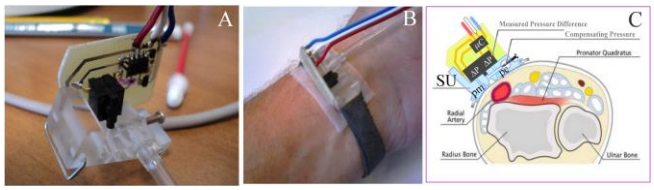

Figure 7. Active sensor for continuous blood pressure monitoring with an air-filled SU camera and without a membrane; its typical arangement on the patient's wrist.

The new construction of ABP monitor undoubtedly fully corresponds to the concept of the Active Sensor. Moreover, it uses the same microcontroller and applications exercising data processing and control were modified slightly. The first experiments gave the same results, which were obtained by the authors earlier and will be described in the next section.

VI. Experiments with the Sensor Developed

The qualitative description of the functioning of the ABP monitor given above could be illustrated by quantitative results obtained by authors in the experiments with the device. Fig. 8 (A) gives the dependence of the SU membrane deformation displacement (in units of the ADC of the displacement indicator SC) on a uniform increase in the pressure inside SU at constant external pressure (the pressure difference at the membrane *P* is the indications of M in mmHg that are counted from the atmospheric pressure). It follows from the plots that in small (±5 mmHg) range of pressure difference on both sides of the membrane the signal of the displacement indicator SC is proportional to it. Fig. 8 (B) presents similar dependences of the displacement of the membrane of SU and pressure *P* inside SU (that is counted from the atmospheric pressure) upon the compensation of uniformly increasing external pressure, with the pressure control *pc* being on. The compensation pressure *P* is created by a pump, when the voltage with pulse-width modulation, which in turn is formed by the microcircuit CD, is applied to its driver PD.

Without specifying the details of the mechanism of formation, note that the resulting *P* is proportional to the control signal *y*, which is directed by the microcontroller μC to CD in response to the membrane displacement ε that is measured by SC (and preceding stored values). The control algorithm was initially chosen as the simplest PID controller [9], with its control signal y_j at the time moment (reading) *j* being formed according to the classic recurrence formula:

$$y_j = y_{j-1} + k_I \varepsilon_j + k_p (\varepsilon_j - \varepsilon_{j-1}) + k_d (\varepsilon_j - 2\varepsilon_{j-1} + \varepsilon_{j-2}), \quad (1)$$

where y_{j-1} is the preceding signal, ε_j, ε_{j-1}, ε_{j-2} are the current and preceding displacements of the membrane from the flat position, and k_I, k_p, k_d are the PID integral, proportional and derivative coefficients, that are selected empirically. It follows from the plots in Fig. 8 (B), that using PID controlling (1) it is possible for a sufficiently wide range of changes of the external pressure (10–170 mmHg) to keep the membrane of SU near the equilibrium position.

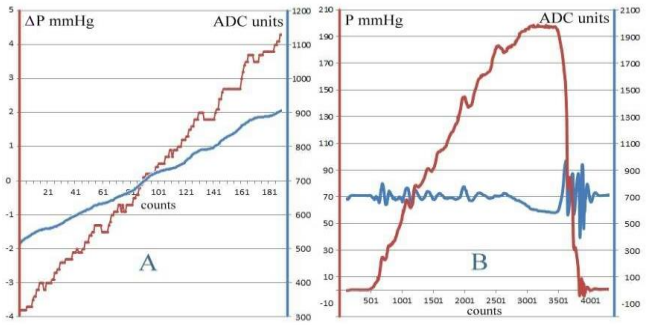

Figure 8. The dynamics of the SU membrane displacement (ADC units) in response to changes of the pressure difference inside and outside the SU.

Based on the calibration data numerous experiments on monitoring blood pressure were carried out. The results of one of the measurements are given in Fig. 9. The lower part of the figure graph shows that the use of the PID control makes it possible to hold the membrane close to undistorted (flat) state for all the time of blood pressure measurement. The quality of such regulation can be estimated by the value of uncompensated difference in pressures equivalent (proportionate) to deviations of the membrane from the flat position (a kind of the membrane jitter).

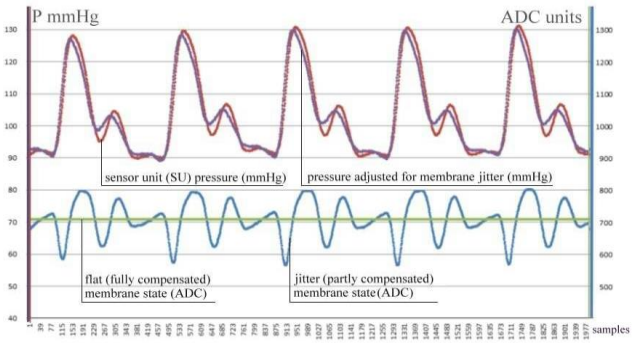

Figure 9. The dynamics of the SU membrane displacement (ADC units) while compensating the time-varying outside blood pressure.

To illustrate the compensation Fig. 9 (top) shows a graph of pressure P inside the chamber SU (the data from pressure gauge M), as well as the same data corrected by the membrane jitter values. In comparison with *P* the uncompensated difference in pressures Δ*P* is not large. Fig. 9 also shows that *P*, on average, coincides with the estimate of ABP, but at the moments of a rapid change in the pulse waveform the controller cannot trace these changes.

The overall impression of the first results of monitoring blood pressure with the help of the above–described active sensor is ambivalent. On the one hand, the sensor does its task – it can be used not only to measure the main parameters of blood pressure – systolic and diastolic blood pressure (range of ABP), but also to observe the varying pulse wave, and not through indirect measurements but by measuring the pressure itself (in mmHg). On the other hand, as it could be initially assumed, a simple PID controller does not provide full compensation, which leads to the distortion of the pulse wave components. Practically, the compensation algorithm based on PID control proved to be very difficult in changing its settings.

VII. Pulse Pressure Compensation Control on the Basis of Pulse Wave Patterns

These shortcomings associated with the incomplete compensation of time–varying blood pressure have a simple explanation.

Firstly, the PID controller used is the special case of linear regulators class and it is well known that linear methods for treating biological signals, especially, linear adaptation methods are used in limited ranges of their changes and only under strictly controlled (for example, laboratory) conditions. This is due to highly non-linear non–equilibrium nature of living systems [3]. Even a small change in physiological state can lead to considerable changes in the result.

Secondly, PID control takes into account only local characteristics of the signals, as it is customary in the theory of dynamic systems. In the case of dynamic systems, such a regulation is natural, as such systems are deterministic. However, living systems and their subsystems are known to be poorly described by models of dynamic systems, even if there is a freedom of choice of corresponding differential equations coefficients. Living systems are much more consistent with models of stochastic, non–deterministic systems.

Thirdly, in solving technical problems low order of PID control is preferred due to its ease of implementation. When somebody deals with a complex biomedical signal, particularly, an ABP signal, low order of control is a drawback.

These facts indicate that the pulse wave of blood pressure is much more like a wideband pulse signal than the sum of harmonic components. In view of the foregoing it can be assumed that a random point process is a closer mathematical model of blood pressure pulsating. A point process is an increasing sequence of time moments (points) of certain homogeneous events, such as the arrival pulse wave, with a random length of time intervals between them. Excluding the changes in the pulse wave pattern, and treating them as a stream of homogenous sequence of events, it would be legitimate to use theoretical methods dealing with the problems associated with the heart rate.

One such well-developed theory today is the theory of radar signal processing. A major problem of this theory is determining the unknown arrival time of an electromagnetic pulse emitted by the radar and reflected from a target. The method of matched filtering should be noted as one of the most effective methods for solving this problem. In this method the filter response is matched with the radar pulse, so that the maximum signal of the filter output is observed at the time, when the reflected pulse arrives. Essentially, the matched filter generates a covariance of sent and received pulses, and it is well known that if both pulses have the same, up to a constant factor, shape, the maximum of the covariance will be achieved when they coincide. The corresponding displacement between pulses will estimate the time of arrival.

The principle of the matched filtering could be interpreted in a different way, more suitable for our tasks. Let us assume at some moment, the time *T* between the ABP pulse wave is known along with the shape of the current pulse, further considering the shape of the next pulse to be the same as the current one, the future of signal can be easily predicted.

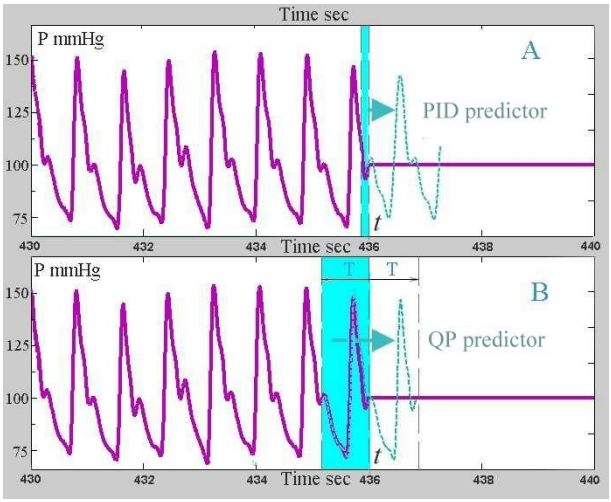

Figure 10. Two methods of predicting the pulse signal future: (A) - by PID regulation and (B) - based on the local quasi-period (QP) estimate.

For this purpose, it should take an existing fragment of the signal of duration *T*, immediately preceding the current time and move it to the time *T* into the future (see Fig. 10). From the theory of the matched filter, it follows that the expected ABP pulse will coincide with the prediction. This idea is illustrated in Fig. 10, which also shows a comparison of the proposed signal prediction with the prediction carried out by PID. As stated above, the PID controller makes a prediction based on current, local estimates of the

fundamental frequency, the amplitude of the main component and amplitudes of neighboring harmonics.

VIII. Conclusion and Future Work

Summing up the results of the investigation it should be noted that for compensatory ABP measurement specifically the current pattern of the pulse wave efficiently predicts the expected signal. This pattern must be dirigible enough to change significantly with changes in the state of the object measured as well as changes in the conditions of its active measurement.

For this reason, realized by the classic regulators including the PID controllers, patterns in the small parametric models of ABP waves are of little use for active blood pressure measurement. An idea of forming the adequate patterns in the task is given in the above qualitative reasoning. It is based on the property of quasi-periodicity of ABP signal.

This property lies in the fact that high variability of the period of heart contractions occurs only at long time intervals, whereas at short time intervals of several seconds or a few heart beats its changes are generally small and fall within a couple of percent.

Therefore, estimating the current period, more precisely the quasi-period T and getting a pulse wave pattern as a signal fragment of T duration immediately preceding the current time moment gives us a good pattern of the pulse wave. Thus, the task of building a pattern for current pulse of the signal is reduced to the task of effective evaluation of its current quasi-period.

The results of developing a new method for monitoring ABP using active sensors and their control based on the pulse wave patterns that have been discussed in this study have demonstrated their high potential in numerous experiments that have been performed by us to test and improve the device developed. This makes it possible for the authors to voice the hope of good prospects of further development up to its adoption in healthcare practice in the very near future.

Acknowledgment

This work was supported by RFBR (Russian Foundation for Basic Research) grant 14-07-00496-a.

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