HeartFelt - Replicable and Accurate ECG Sensor Architecture for Real-Time Remote Monitoring

Daniel Rosner, Dumitru-Cristian Trancă, Răzvan Tătăroiu, Adrian Cristian Petrescu
{daniel.rosner, cristian.tranca, razvan.tataroiu, adrian.petrescu}@cs.pub.ro
Computer Science Department
University POLITEHNICA of Bucharest
Bucharest, Romania

Dan Iorga
iorga.dan@gmail.com
Faculty Electrical Engineering, Mathematics and Computer Science, TU Delft
Delft, Netherlands

***Abstract*— Heart affections are currently the main type of illness affecting the elderly, and pose unique difficulties both in diagnostic and in treatment effects tracking and long term evolution monitoring. The already stretched out medical system cannot handle the requirements of long term monitoring of patients with severe heart conditions, and thus, automated long term monitoring solutions are required. The present work displays a practical analysis of the hardware and software solutions available for ambulatory ECG (electrocardiogram) monitoring, by proposing a replicable ECG sensor architecture - HeartFelt. The proposal is detailed and validated by means of a complete hardware and software implementation, confirmed by tests performed in cooperation with medical staff. The exposed architecture offers the possibility of long term monitoring of an ECG signal from a patient in ambulatory environment, and offers a balance between precision, ergonomics, autonomy and productions costs.**

Keywords—ECG, monitoring architecture, heart, remote-monitoring, medical system, automatic processing

I. INTRODUCTION

Tele-medicine has known a large expansion of both application areas and degree of penetration in the medical systems in developed countries over the last decades [1], [2]. This expansion in usage is mainly driven by the need for cost-effective solutions in medical systems that have to care for a constantly increasing number of patients while having limited resources, both material, but especially human resources.

Medical systems in developed countries are approaching a critical state, derived from the following conditions:

- A constant rise in life expectancy at birth, mainly due to increase in quality of life and to advances in medicine [3]. According to [4], which deals with statistics for countries inside the Organization for Economic Co-operation and Development, the average life expectancy at birth has risen by 10 years inside the 1970 – 2010 period;

- A constant increase in life expectancy for elderly people suffering from illnesses that would have been fatal a few decades ago;

- Higher costs of new treatments and medicines that are constantly being developed. According to [5], healthcare expenditures in 2012 reached an average of 9.6% of GDP, while in the USA these are peaking at 17.6%;

- An aging population leading to a lower ratio of active (people who work and finance the medical system) to non-active population (mostly children and elders) [3].

As such, one of the major challenges faced by the health care system is to provide quality healthcare services to an increasing number of elder patients suffering from affections that require long-term supervision, while using limited financial and human resources.

Research such as [6], [7] indicate heart-related problems as one of the main conditions that require long term monitoring in elder patients, while [8], [9] indicate ECG measurements as one of the most important vital signs to be measured by telemedicine systems. Solutions for long-term at-home supervision of vital signs for elder patients are needed in order to both reduce pressure on the medical system, and to increase both the accuracy of the offered treatments and the living quality for elderly patients.

The contributions of this work are: a complete and comprehensive guide for a wearable ECG monitoring architecture; an open-source, easy to interface, low-power, low-cost implementation, accessible to researchers and hobbyists that want to build a scalable and adaptable monitoring solution.

II. RELATED WORK

In [10] the authors describe the general guideline for building a low cost ECG system. The study emphasizes the need for such systems, motivated by the ever greater pressure exercise on the current medical system, the high percentage of patients with cardiac problems among the elderly, as well as the economic opportunity provided by a dynamic market. The article recommends the use of a 3-lead system, employing an instrumentation amplifier (INA) and a series of filters for the analog front end (AFE). It also suggests various ADCs for signal acquisition and microcontrollers for the central processing unit.

In [11] the authors propose a development platform for ECG signal acquisition, also based on a three-lead system. Within the AFE, a 60Hz notch filter is used for eliminating power-line induced noise. The authors also underline the importance of using a right-leg drive (RLD) electrode that provides common-mode negative feedback. The bio-signal is amplified by an INA with a very high CMRR (common-mode rejection ratio) and input impedance. The signal is then passed through a 150Hz low-pass filter, as well as a 60Hz notch filter. A 12bit ADC samples the signal at 400Hz. For transmission, an 802.11b solution is used. Overall, the solution offers good measurements, but with costly parts and substantial power requirements and size.

In [12] the authors present a solution for ECG monitoring based on a ZigBee platform for wireless communication. The authors propose a hardware solution for measuring the ECG signal inside a 0.5 - 35Hz bandwidth, that is considered enough for monitoring applications (as opposed to diagnosis application that require 0.05 to 150Hz bandwidth). For the AFE, the authors use an INA with gain of 50 together with a RLD and an integrator used to minimize interference and baseline wander. A low-pass active filter is used to reject high frequency noise and to further amplify the signal up to 800 times. A sampling rate of 100Hz was chosen for the 10bit ADC used.

Liu et al. propose a multiple parameter system that deals with ECG, respiratory rate, motion sensors and temperature. [13]. For the ECG sub-system, they used a dedicated ADS1292 front-end that has built in respiratory rate detection.

Ma et al. developed another multi-parameter monitoring system that includes an ECG monitoring sub-system, with a proprietary implementation [14].

The HEARTRONIC project proposed by Rocha et al. is focused on early stage detection of heart related problems [15]. Furthermore, the authors propose a system that can issue early warnings on soon-to-happen (2-3 hours range) heart conditions.

The authors of the current paper have previously proposed a simpler hardware approach for ECG signal acquisition [16].

III. A MEDICAL OVERVIEW

The cardiac activity generates an electric field that can be detected on the body by means of electrodes placed on the patient's skin. These electrodes will pick up cycling voltage difference wave-form that is referred to as ECG.

In order to measure a patient's ECG, the measuring device is connected by means of electrodes that may be attached in various parts of the body. There are three ways of connecting the electrodes: direct - inside the cardiac cavity, semi-direct (esophagus), and indirect – on the skin. This last method is non-intrusive and is the most common one. Particularly, in long-term monitoring, it is the standard method.

There are three ways of connecting the electrodes on the body of a patient:

- **Standard or bipolar deviations** – obtained by connecting two electrodes on two regions equally distant from the heart: right forearm, left forearm, left feet.

- **Unipolar deviations** are obtained by connecting three electrode on the above mention body members, and a forth one as far away from the heart as possible.

- **Precordial derivations** are unipolar deviations that register potential differences form myocardial regions beneath the exploratory electrode. The amplitude of the deflections is larger than in the first two cases, as the main electrode is closer to the heart.

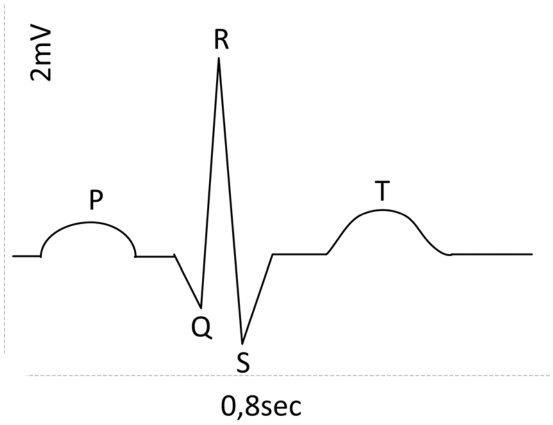

Fig. 1: Normal GSR shape and typical values

In most ambulatory remote monitoring systems, the bipolar deviations are used, as this can yield a good image of the patients ECG, while measurable even when the patient is moving. However, this requires noise cancelation that can be obtained by feeding a negative feedback signal to the patient's right leg, through a third electrode (the most distant external point from the heart). This configuration is also preferred as it is less intrusive and easy to wear.

Normal electrocardiogram presents a series of waves, segments and intervals. The deflections are named P, QRS and T. The segments are portions between the waves, and intervals are portions the comprise waves and segments.

IV. HARDWARE APPROACH AND IMPLEMENTATION

The hardware design for the HeartFelt architecture is optimized for mobility, low power and high tolerance to electromagnetic interference. A block diagram of the system hardware is depicted in Figure 2.

As some diagnosis methods require correlation between ECG and other vital signs, the system can also samples acceleration, temperature, and pulse and oxygen saturation data from the respective subsystems and sends it wirelessly to the monitoring station. It consists of a low-power ATmega324PA microcontroller and associated support circuitry.

The ECG subsystem performs signal conditioning and acquisition with regards to monitoring the electrical heart function. It consists of an analog front-end (AFE) which performs signal amplification, noise rejection and filtering, and an analog-to-digital converter (ADC). A discrete design was chosen for the ECG subsystem due to its flexibility and ease of debugging.

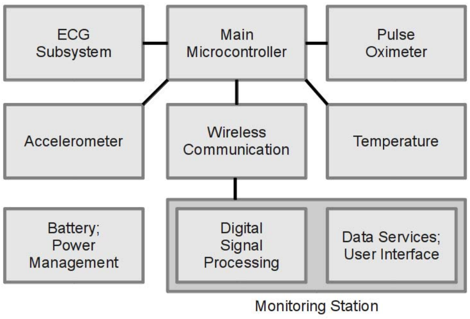

Fig. 2: System block diagram

The purpose of the AFE (Figure 3) is adapting the low-level ECG signal, which is in the order of millivolts, to the ADC input range of hundreds of millivolts, while attenuating parasitic signals such as those originating from power lines and radio transmissions.

The AFE also rejects various low-frequency parasitic signals originating from the electrode-patient connection. It employs a high-performance instrumentation amplifier (INA), a right-leg-drive amplifier, an active filter an output amplifier. The INA is based on a standard three-operational-amplifier integrated circuit with additional radio-frequency rejection input filters. It amplifies the potential difference between the measuring electrodes, which represents the useful ECG signal, while rejecting the common-mode noise.

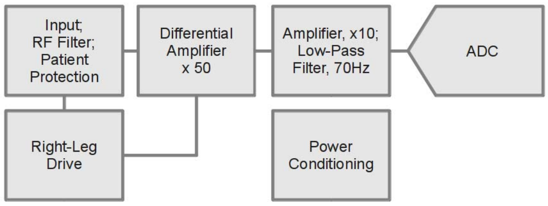

Fig. 3: AFE block diagram

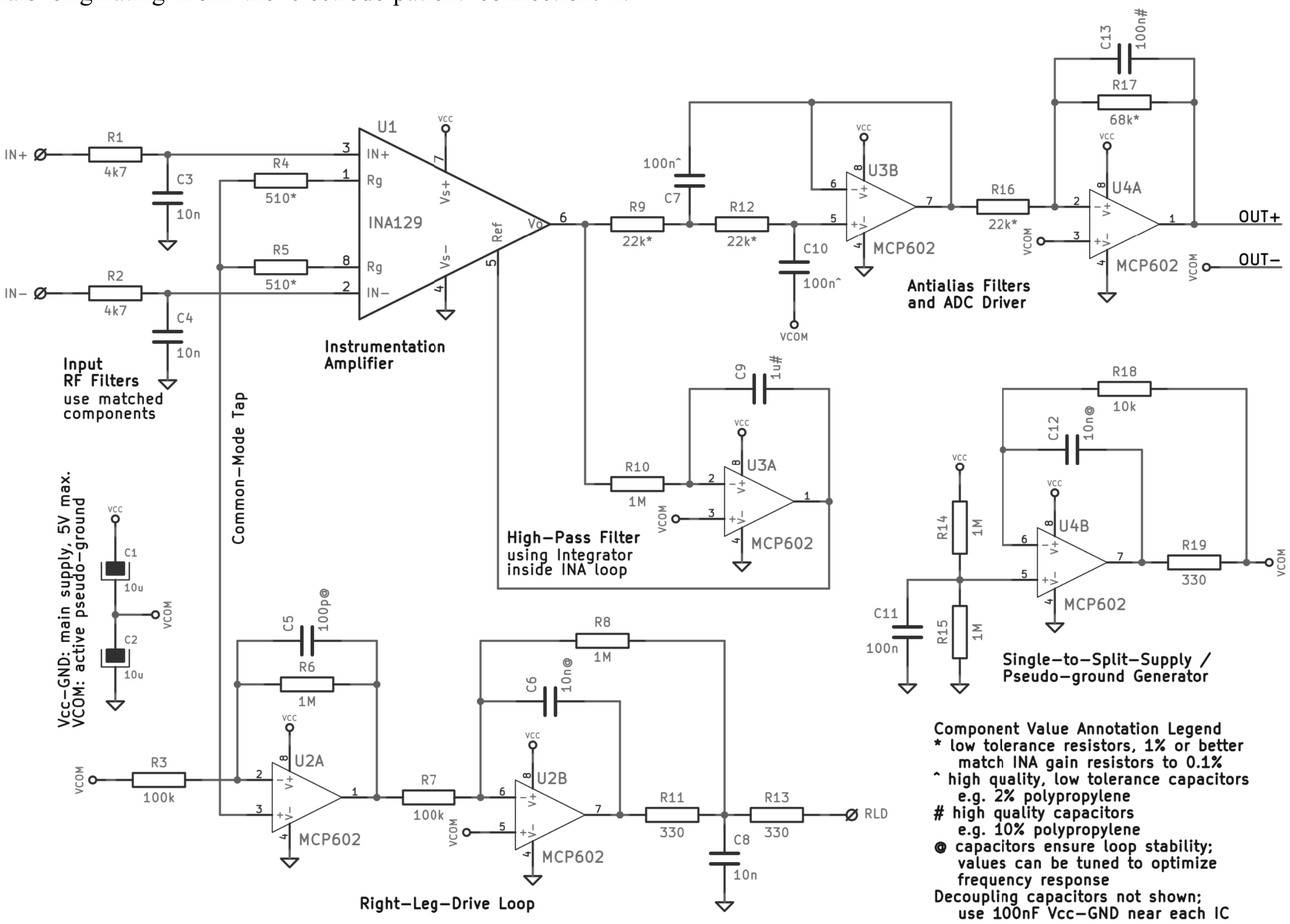

Fig. 4: Schematic of the hardware implementation for the AFE

The integrated circuit is designed to have its differential gain set by a single external resistor. The gain in this instance is set to approximately 50, while the resistor is split into two equal halves in order to generate a common-mode signal at the middle. This common-mode signal, which is interfering and must be rejected, is fed into the inverting right-leg-drive amplifier, which forms a control loop through the patient's body with the purpose of reducing said signal. The right-leg-drive amplifier is fitted with an output radio-frequency rejection filter, as well as an appropriate frequency-compensation components to ensure closed-loop stability. An integrator is also placed in the feedback loop of the INA, on the positive reference input, yielding a high-pass filter response (0.5Hz). This attenuates unwanted signals related to the contact potential of the electrodes. The detailed AFE implementation is presented in figure 4.

The ECG signal is then fed into a second-order low-pass active filter employing the Sallen-Key topology. This filter rejects high-frequency noise and power-line interference. Ideally, this filter would render 50/60 Hz noise undetectable. In practice however, such a filter would require expensive precision components and, more importantly, additional power consumption and circuit board space. As this is a low-power, mobile design, the present hardware design only provides moderate attenuation at power-line frequencies, delegating its implementation to digital filters implemented in the main microcontroller software or on the monitoring station.

Even so, oscilloscope testing reveals that 50Hz analog noise is below 20% of the useful analog signal. The harmonics of the power-line noise were also taken into consideration: as well as being considerably lower in amplitude than the fundamental, they are also more strongly attenuated by the analog filter as well as the digital filter thanks to a relatively high sample rate.

Following the active filter, an output amplifier feeds the signal to the ADC, while providing an additional first order low-pass filter (150Hz).

The entire signal path (INA, filter, output amplifier) is designed to work with a single low-voltage DC power supply, such as a rechargeable battery. In this regard, a pseudo-ground at half the power supply voltage is generated, allowing operational amplifiers to work in a standard dual-supply topology down to DC. This also takes advantage of the ADC integrated circuit having a bipolar differential input, allowing both positive and negative voltages with regard to the pseudo-ground to be measured.

The accelerometer module translates the three orthogonal acceleration components to analog voltages which are directly sampled by the main microcontroller.

The temperature module consists of an integrated circuit temperature sensor which connects to the main microcontroller by means of a standard I2C interface.

The wireless communication module sends the acquired data to the monitoring station. It is currently implemented as a Bluetooth module which is interfaced to the main microcontroller via a serial UART interface and an auxiliary connection-established signal.

Bluetooth was chosen from three considerations: it is compatible with a wide range of smart-phones, laptops and other electronic devices; it supports low-power mode for close range communication (e.g.: with a mobile-phone in the patients pocket); studies as [17], [18], show Bluetooth to be an excellent choice for wireless transmission in medical applications. Moreover, as Bluetooth is a technology currently integrated in numerous modern equipment, it provides an effective solution for fast integration in other monitoring and assistance systems.

V. SOFTWARE APPROACH AND SOLUTION

The HeartFelt software architecture is optimized to offer excellent signal processing while keeping computational requirements low.

A. Signal Filtering

Noise induced by power-line interference, the patient's movement or by other signals present in the human body can distort the ECG signal. These elements have to be first removed before any effective signal analysis can take place. Performing these operations entirely in the analog hardware section would significantly increase complexity, price and power dissipation; eliminating all artifacts in software is not entirely feasible and would in any case require powerful computing hardware; the current architecture uses an analog high-pass filter and anti-aliasing filter, combined with software processing, yielding a reasonably clean signal on which analysis can be performed.

B. Baseline wander removal

A strong variation of the signal baseline can occur due to breathing, chemical contact potential and other factors. A high-pass filter is used to attenuate this baseline wander. The cutoff frequency of the filter can be safely chosen between 0.5Hz and 1Hz without eliminating any important signal information. The high-pass filter was implemented in software using the "forward-backward IIR filtering" technique. It consists of a series of steps that can be described by the following equations:

$$\begin{gathered} y_1[n] = h[n] * x[n] \\ y_2[n] = y_1[-n] * h[n] \\ y_3[n] = y_2[-n] \end{gathered} \tag{1}$$

This filter produces no phase distortions; however, it cannot be implemented in real time. To overcome this, the signal can be delayed for 3 to 4 seconds, thus obtaining a pseudo-real time signal. Studies indicate this to be acceptable, even in real time monitoring systems [6].

C. Power-line interference removal

Power-line interference at the fundamental frequency of 50/60 Hz is a common problem with low voltage signals, and can be overcome in many different ways. For example, a stop-band Butterworth filter is very effective and easy to implement but can introduce distortions in the signal.

A solution is proposed in [19] consisting of a filter that extracts a sinusoid at 50/60Hz from the signal and adapts the amplitude of the sinusoid at every new sample. We consider the following signal that is subtracted from the signal.

$$v[n] = \omega_0 \sin(\omega_0 n) \tag{2}$$

The sinusoid is generated by the following equation:

$$v[n] = 2cos\omega_0 v[n-1] - v[n-2] + u[n] \quad (3)$$

where v[-1]=v[-2]=0.

At every step the amplitude of the sinusoid has to be adjusted.

$$\begin{gathered} e[n]=x[n]-v[n] \\ v'[n]=v[n]+\alpha\, sgn(e'[n]) \end{gathered} \quad (4)$$

The value of α determines how fast the filter converges.

The final step is to subtract the sinusoid from the signal:

$$y[n] = x[n] - v'[n] \quad (5)$$

D. High- frequency noise removal

The previous filters eliminate the most significant artifacts that can be found within a signal but there is usually some high frequency noise remaining. A low-pass filter can be implemented, but this can affect the shape of the R peaks.

The Savitzky-Golay filter [20] was used as an alternative. This is a smoothing filter that approximates a local polynomial regression through a FIR filter.

The coefficients for the filter can be calculated with software tools such as Matlab. A 7th order polynomial and a frame size of 25 were used.

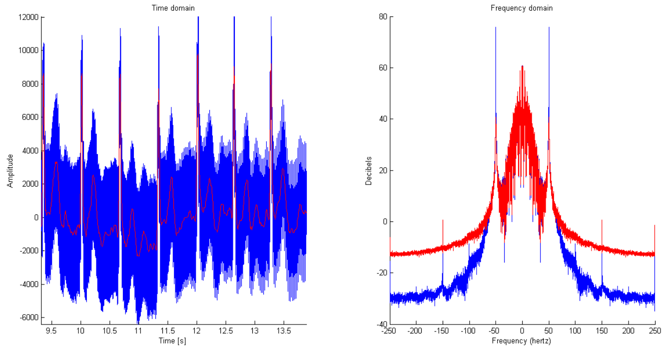

Fig. 5: Time domain (left) and Frequency domain (right) for the original signal (BLUE) and the filtered signal (RED)

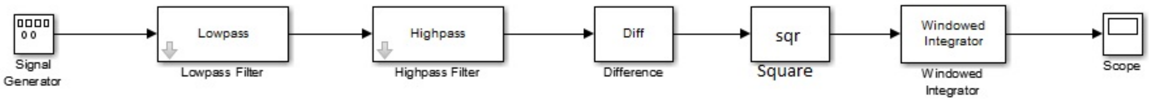

Fig. 6: The filters that constitute the Pan-Tomkins algorithm

E. Interval detection

To evaluate a patient's health status a few intervals have to be measured. The easiest way to do this is to first determine the position of the QRS complex then based on this, the position of the rest of the components.

The algorithm described in [21] can be easily implemented and provides a good approximation of the position of the QRS complex. First the signal passes through a band-pass filter that attenuates the noise. It is then differentiated, squared, and integrated over 150ms. The output consists of a series of peaks that indicate the position of the QRS complex.

Knowing that the QRS complex has the highest frequency in the signal, the STFT (Short-Time Fourier Transform) can be efficiently used to determine the position.

One of the most accurate methods to determine the QRS complex is to use the wavelet transform. Compared to the STFT, the wavelet transform does not require a fixed window size and is more precise in determining the exact position.

There are a large number of types of wavelets that can be used with different results but their comparison is beyond the scope of this paper. For a detailed review of these wavelets there are articles such as [22]. Good results were obtained using the Daubechies – see Figure 7.

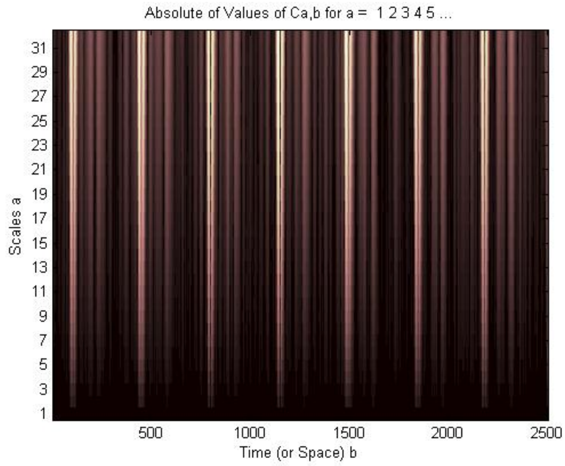

Fig. 7: Wavelet transform of the ECG signal.

Searching for the maximum in a small window to the left of the QRS complex, the location of the P wave is found; searching in a small window to the right, the location of the T wave is found. The size of the window is chosen based on the knowledge of these values in healthy patients.

VI. RESULTS

Figure 8 illustrates the accurate identification of the positions for the P, Q, R, S, and T waves, which validates the proposed implementation.

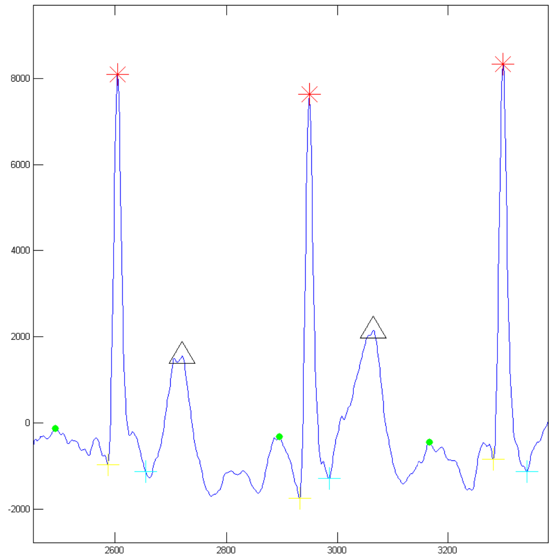

Fig. 8: The positions of the P, Q, R, S, and T waves

VII. 6. Conclusions

The current research offers a concise but comprehensive evaluation of current solutions for ECG monitoring, as well as effective architecture for developing a complete mobile ECG monitoring system.

The HeartFelt architecture is validated by a physical implementation that balances requirements for portability and autonomy with implementation costs. The hardware design for this solution also adds 3 auxiliary parameter measurements, which provide a better insight into the patient's health status, without significantly impacting autonomy or cost. The software component of the validation implementation offers excellent results with minimum computational consumption.

The added wireless data transmission system also allows the physical parameters monitoring system to seamlessly integrate with complex monitoring and quality of life assistance systems.

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