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Wireless Device for Patient Monitoring

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Abstract— **The increasing request of patients, suffering of chronic diseases, who wish to stay at home rather then in a hospital and also the increasing need of homecare monitoring for elderly people, have lead to a high demand of wearable medical devices. Also, extended patient monitoring during normal activity has become a very important target. Low power consumption is essential in continuously monitoring of vital-signs and can be achieved combining very high storage capacity, wireless communication, and ultra-low power circuits together with firmware management of power consumption. This approach allows the patient to move unconstraint around an area, city or country. In this paper the design of ultra low power wireless monitoring devices based on ultra low power circuits, high storage memory flash, bluetooth communication and the firmware for the management of the monitoring device are presented.**

Index Terms—**healthcare, telemedicine, low power microcontroller, wearable device.**

I. Introduction

Phenomena of ageing population observed in most developed countries [1] and prevalence of chronic diseases have increased the need for chronic and geriatric care at home [2]. The task of patient monitoring may be achieved by telemedicine (enabling medical information-exchange as the support to distant-decision-making) and telemonitoring (enabling simultaneous distant-monitoring of a patient and his vital functions) both having many advantages over traditional practice. Doctors can receive information that has a longer time span than a patient's normal stay in a hospital and this information has great long-term effects on home health care, including reduced expenses for healthcare.

Despite the increased interest in this area, a significant gap remains between existing sensor network designs and the requirements of medical monitoring. Most telemonitoring networks are intended for deployments of stationary devices that transmit acquired data at low data rates and rather high power consumption. By outfitting patients with wireless, wearable vital sign devices, collecting detailed real-time data on physiological status can be greatly simplified. But, the most important effect is the widely social integration of peoples with disabilities or health problems that need discreet and permanent monitoring.

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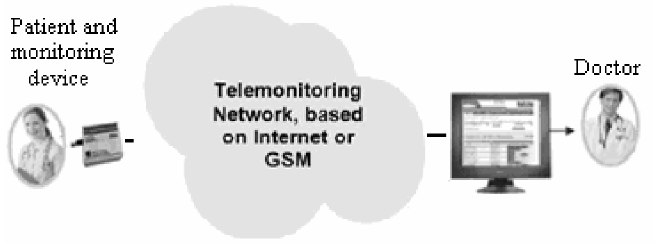

Fig. 1. The general structure of telemonitoring network

It is well known that Bluetooth represents a reliable and easy solution for signal transmission from a portable, standalone unit to a nearby computer or PDA [3]. When a real time rudimentary analysis program detects pathological abnormality in the recorded data an alert is sent to the telemonitoring centre via internet or GSM/GPRS using a PDA or computer.

The needs of the society and market trend suggest that healthcare-related applications will be developed significantly. Particularly, PDA phones interfaced with wearable sensors have an enormous market as they will enlarge the goal of healthcare and mobile devices [4].

The development of ultra low power wearable monitoring device unit is propounded. The main functions of the wearable device consist in signal acquisition, rudimentary processing, local data storage and transmission to the remote care centre. In Fig.1, the general architecture of the telemonitoring network is presented.

This paper presents the initial results and our experiences for a prototype medical wearable device for patient monitoring taking into account some hardware and software aspects regarding vital signs measurement, robustness, reliability, power consumption, data rate, security and integration in a telemonitoring network.

II. Wireless monitoring device

The monitoring device is build using custom developed hardware and application software. The block diagram is presented in Fig. 2 [7]. Low power amplifiers and sensor are connected to the device [6], for vital parameters acquisition. The health parameters acquired are: heart rate, heart rhythm regularity, respiratory rate, oxygen saturation and body temperature. The monitoring device includes also a custom made 3-lead ECG amplifier.

The digital board includes an ultra low power microcontroller and interface circuits for keyboard, SD/MMC card and Bluetooth. The low power sensors for the above mentioned parameters are connected to the digital board using digital port or specific converters.

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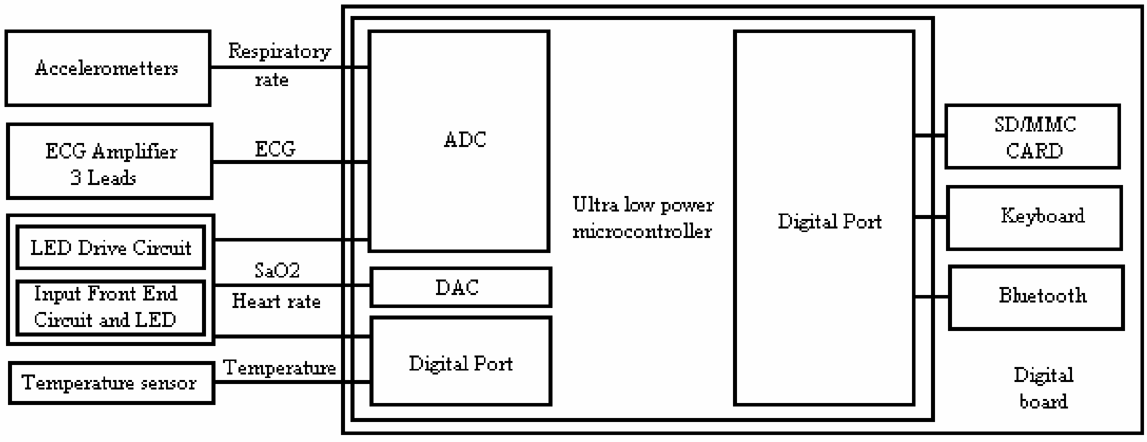

Fig. 2. Block diagram of the monitoring device

The wireless communication via Bluetooth with a computer or PDA and the flash memory for raw data information storage are two of the most important facilities of the proposed wearable device.

The software application running on computer and PDA assures the communication via internet respectively GSM/GPRS [3], [5].

The monitoring device is built using an ultra low power microcontroller (MSP430 from Texas Instruments) that has a 16 bit RISC core with clock rates up to 8MHz. An important advantage of this family of microcontrollers is the high integration of some circuits that offer the possibility to design devices with open architecture: digital ports, analog to digital converter (ADC) and digital to analog converter (DAC). MSP430 microcontroller also has a built in hardware multiplier which is a valuable resource for digital filter implementation. The custom made electronic boards used for data acquisition are all designed with low power circuits. The system is modular: every board is detachable.

The on board storage device (SD/MMC card) with FAT32 file system is used for raw data recording together with signals processing results. The radio module Bluetooth [8] designed for short-range communication is used for data transmission between monitoring device and PC or PDA. This module has low power consumption being used only when data transfer is performed.

III. Functionality

The ultra low power wireless monitoring device is able to acquire simultaneously the physiological parameters mentioned in the previous section and also to perform rudimentary digital signal processing on board. The signals are continuously recorded in separate files on flash memory for feature analysis. Once pathological abnormality is detected, the monitoring device requests a transmission trough PC or PDA to the remote care centre.

For the respiratory rate it is used a low power three axis low-g accelerometer MMA7260QT [13]. This is a low cost capacitive micromachined accelerometer that features signal conditioning, a 1-pole low pass filter, temperature compensation and g-Select which allows for the selection among 4 sensitivities.

The accelerometer outputs are read and processed once at 10 ms (100 times per second).

Because this is a low power system, the accelerometer is disabled between two conversion cycles. Each conversion cycle has a 2 ms startup period for accelerometers to recover from sleep mode.

Due to the sensitivity of the accelerometer and the high resolution of analog to digital converter, the useful signal (the output voltages of accelerometer) has a large amount of superposed noise with possible sources in body moving or even heart beating. In Fig. 3, the acquired signal affected by noise is presented. There can be seen the initial rise of voltage corresponding to the start of chest movement. This is explained by the fact that the movable central mass of g cell moves in opposite direction of the movement. As the chest movement velocity becomes constant, the acceleration decreases and movable central mass comes back to initial position. This settling process implies acceleration opposite to the direction of movement.

Noise filtering is achieved by using a 128 point rolling average filter. Mathlab simulation was used with real datasets for filter calibration. Without filtering, the signal is almost unusable for further processing. After filtering, the noise is substantially reduced.

To easily distinguish between positive and negative accelerations (e.g. to determine the sense of movement) the accelerometer has an offset of half of supply voltage for 0g acceleration. So, for positive or negative accelerations the corresponding output voltage is positive in all cases.

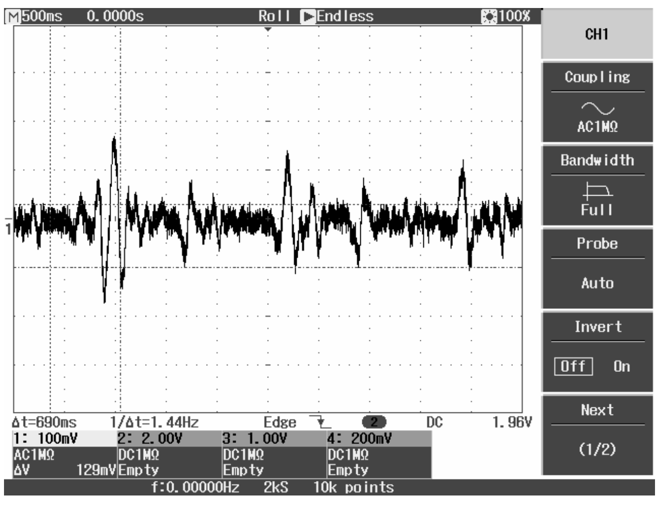

Fig. 3. The accelerometer time response

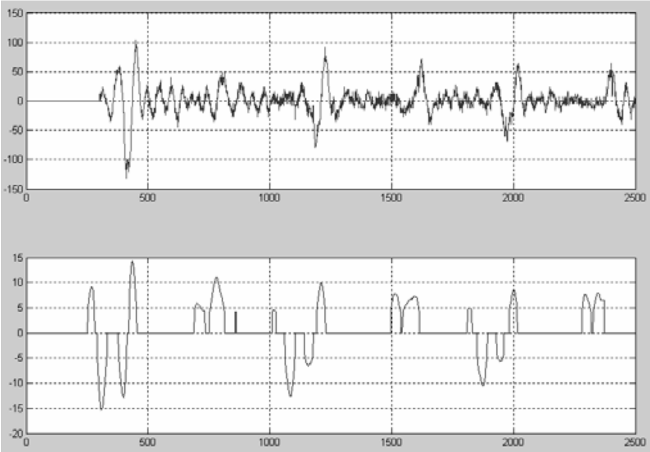

Fig. 4. The processed signals

To make the difference between positive and negative accelerations, it is necessary to fix the reference value corresponding for 0g. This is done at the application startup, when an average with 512 points is made.

In Fig. 4, the results of the algorithm are plotted. The first plot is the raw unfiltered data from accelerometers outputs in mV. This plot is a 2500 points record of accelerometer output for a time period of 5 seconds. The second plot represents the filtered input acceleration.

The custom made ECG module uses three leads. The ECG signals are acquired using micro power instrumentation amplifiers and micro power operational amplifiers. Both are single power supply. A very important feature of the instrumentation amplifier is that it can be shutdown with a quiescent current of less than 1μA. Returning to normal operations within microseconds, the shutdown feature makes is optimal for low-power battery or multiplexing applications [10]. The propounded functioning is: on power on the instrumentation amplifier is shutdown. With 10 μs before a conversion cycle starts the instrumentation amplifier is returned from shutdown mode and back again when the conversion cycle finishes.

The temperature is monitored every minute using an ultra low power temperature sensor.

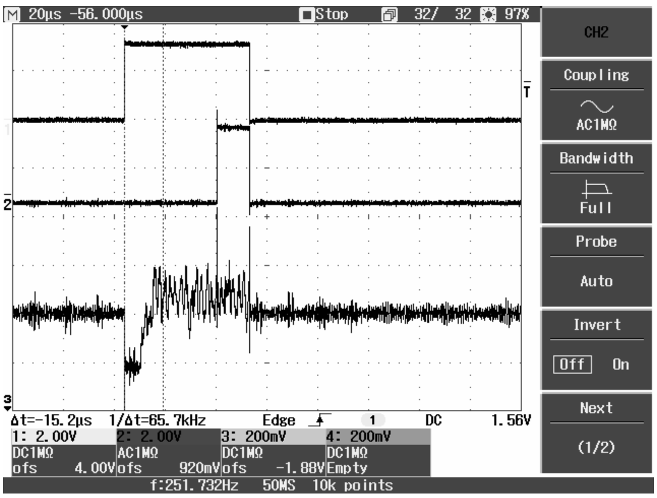

Fig. 5. The instrumentation amplifier output

Meanwhile, the sensor is shutdown saving maximum power by shutting down all device circuitry other then the serial interface, reducing current consumption to typically less than 0.5 μA [9].

Because the read/write operation with flash card can reach a current consumption of max 60mA, these are done with blocks of data [11]. The file system on flash card is FAT32 witch gives the possibility of reading those files with any PC with a card reader.

The average current consumption of Bluetooth circuit is around of 35 mA. For this reason, the Bluetooth communication is used in two ways: always connected transmitting live or stored data (signal processing and data storing are not performed) and not connected, the circuit is in sleep mode (signal processing and data storing are performed). If a critical situation occurs the circuit is awaken, the communication starts, and in the same time data storage is performed. When files are downloaded every monitoring operation is stopped.

In the case of blood oxygenation measurement, for driving and controlling the LEDs a solution proposed by Texas Instruments was used[14]. It is based on two LEDs, one for the visible red wavelength and another for the infrared wavelength. The photo-diode generates a current from the received light. This current signal is amplified by a trans-impedance amplifier. OA0 (microcontroller built in), one of the three built in op-amps, is used to amplify this signal. Since the current signal is very small, it is important for this amplifier to have a low drift current [15].

Because of the high level of analog circuits' integration the external components involved in hardware development are very few. Furthermore, by keeping the LEDs ON for a short time and power cycling the two light sources, the power consumption is reduced [15].

IV. FIRMWARE ISSUE

The developed firmware consists of several tasks and each of them manages a particular resource as presented in Fig. 6. The communication between tasks is implemented with semaphores and waiting queues allowing a high level of parallelism between processes. Each process may be individually enabled or disabled. This feature is very important in increasing the flexibility of the application: if real time monitoring is desired, then SD Card Process may be disabled and Bluetooth Process is enabled, if only long term monitoring is desired then SD Card Process is enabled and Bluetooth Process may be disabled. This has a positive impact on power consumption because only the resources that are needed are enabled for use.

A logical cycle of the device operation is presented in Fig. 6. In the first step (thread), a buffer of data is acquired. The ADC module works in sequence mode and acquires data from several channels at a time. The ADC channels are associated with temperature, accelerometer, ECG signals and SaO2 inputs of electronic interface modules. One sequence of ADC conversion is made by one sample for each of these inputs. The data is formatted and then stored in a buffer. Only when the buffer is full the following thread may begin. Depending on application requirements the storage, transmission or analyze of the data may be performed.

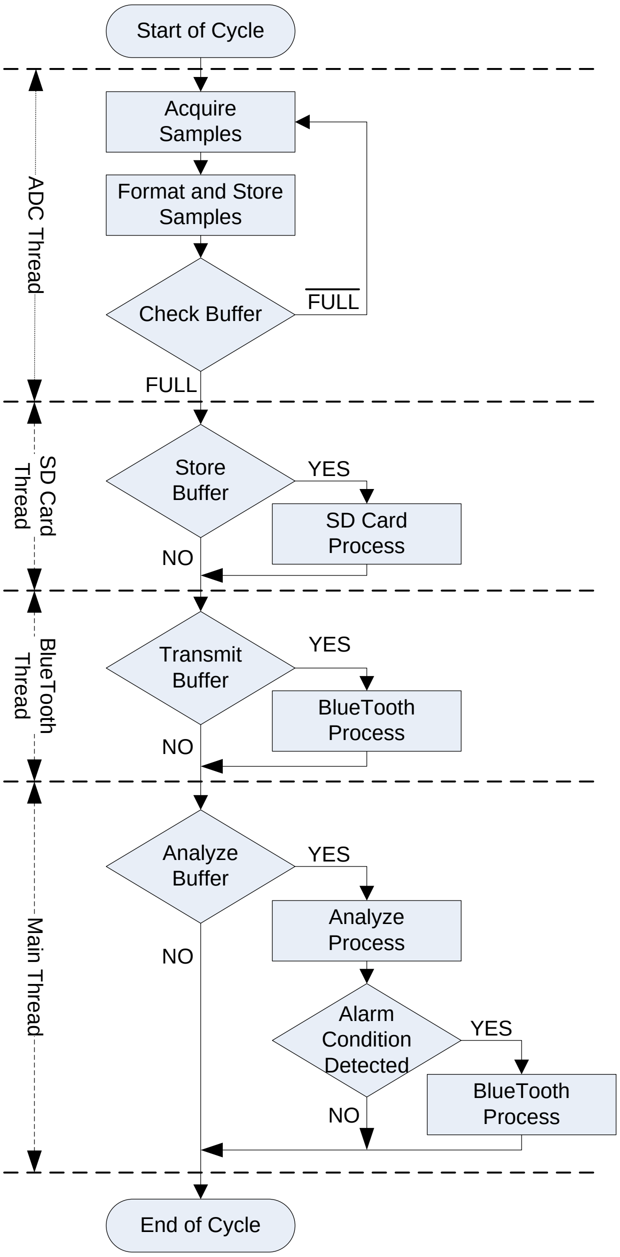

Fig. 6. The tasks cycle

Encryption of personal data is provided. Some files that store personal data related to the patient are encrypted and can be accessed only using the proper decryption algorithm. Only a part of the files are encrypted because there must be kept a balance between power consumption and computing power requirements. For space saving, a light compression algorithm may be activated as an extra feature of the device. The activation of this feature has a negative impact on the overall power consumption.

The communication process implements an application layer protocol for data exchange between the device and other Bluetooth enabled devices. The communication requirements must be limited to a minimum rate in order to save power. The Bluetooth module is powered on when a message has to be sent. The message transmission rate is kept low by using internal buffering and burst communication. The Bluetooth module can be also waked up by an incoming message that may embed commands to the device.

The analysis process implements some rudimentary signal processing tasks in order to detect anomalies in physiological activities. Only few vital parameters anomalies are locally detected (by the wearable device) and trigger an alarm event. More complex signal processing and dangerous condition detection are implemented on a remote computer which has the required computing power. The remote computer receives raw data from the device and does the required analysis.

The physical processes that are monitored have a slow changing rate leading to a low sampling rate requirement. This allows the controller to be in sleep mode for an important percentage of operating time.

The acquisition of one full data buffer takes the most time of the operation cycle. The time needed for the execution of each remaining thread is shorter compared to the ADC process.

A double buffering scheme is used: meanwhile a buffer is filled up with new data samples, the second buffer may be processed (stored on SD card, transmitted via Bluetooth or locally analyzed in order to detect conditions that should trigger alarm events). When the new data buffer is acquired the buffers change (simple pointer assignment) occurs: the previous processed buffer becomes empty, the acquired data will be placed in it and the actual filled buffer will be processed.

The device must be reliable. For this purpose a power monitoring function is designed. To implement this feature the ADC module is used. One of its inputs monitors the voltage level across the batteries. The alarm events are designed to preserve as much power as possible. If a voltage threshold is reached then an alarm event is triggered and a resource could be disabled or its usage restricted only if configured so. For example, the messages content and its transmission period could be modified in order to save power but also to ensure a safe operation.

The computer software (see Fig 7) is a MDI application (Multiple Document Interfaces). A connection can be established with more than one device (each one has a unique id). The software is always listening for connection with registered Bluetooth monitoring device. This means that before data transmission, the device id has to be known by the software. This is done by entering the code manually, or by answering *yes*, when the application prompts for a communication request. The ID is stored for further use. The communication session is established in two ways: by the device when a critical situation occurs or by the computer when a real time monitoring or file download is requested. This is done by sending a break condition to the device to wake it from the deep sleep state [9]. The software assures the connection to a server via Internet and also data transmission.

Fig. 7. Software user interface

V. CONCLUSION

The work of this paper focuses on design and implementation of an ultra low power wearable device able to acquire patient vital parameters, causing minimal discomfort and allowing high mobility. The proposed system could be used as a warning system for monitoring during normal activity or physical exercises. The active collaboration with The Faculty of Biomedical Engineering from University of Medicine and Pharmacy from Iasi (Romania) and with several hospitals offered the opportunity to test the prototype. Preliminary results were so far satisfactory. This wearable device will be integrated into TELEMON healthcare systems to provide real-time vital parameters monitoring and alarms.

In conclusion, this paper presents some challenges of hardware and software design for medical wearable device based on low-power medical sensors and microcontroller with a recent tremendous impact in many medical applications. Obviously, the results demonstrate that there is still significant work to be done if the wearable device is effectively integrated in a large network of medical sensors.

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