ORIGINAL PAPER

MOBILE CARDIAC TELEMETRY SYSTEM FOR ISOLATED IMMUNOSUPPRESSED PATIENTS

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Abstract. *This article represents an experiment for a portable open-source ECG monitoring system, purposed for isolated immunosuppressed patients, with heart conditions. At the current stage of development, the proposed experimental MCT (Mobile Cardiac Telemetry) system can be wirelessly used in conjunction with a smart phone and a real-time graph streaming application, through the BLE (Bluetooth Low Energy) module of an IoT embedded platform. At the moment, the system is intended for autonomous low power consumption and short wireless proximity range suited for close monitoring of patients by themselves at home, or by the medical staff in the isolated hospital ward rooms or mobile isolation containters.*

Keywords: *ECG; MCT; wireless monitoring; heart disease; immunosuppressed patients.*

1. INTRODUCTION

The electrocardiogram (ECG or EKG), is one of the simplest, common, and oldest, heart investigation technique still available today. Of course there are more advanced ways for heart diagnosis, such as echocardiogram, cardiac CT scan (Computerized Tomography) or cardiac MRI (Magnetic Resonance Imaging) [1], but the main focus of the article is on the ECG portability and power consumption functions [2].

Although the standard ECG diagnostic systems provide a brief chart of the cardiac electrical activity for more than 10 seconds time, the ambulatory ECG systems can provide data over an extended period of time from days to years, allowing the user to capture and store more consistent information over the abnormal cardiac events and full activity of the cardiac muscle [3].

The commercial, wearable, ambulatory monitoring devices, can be quite useful for basic usage [4], to measure biometric parameters such as temperature, bpm (heart rate), ECG, SpO_2 (oxygen saturation), GI (glycemic index) and many other essential health parameters, without the need of a doctor and hospitalisation, thus preventing the spread of contagious viruses, bacteria or getting various nosocomial infections found in the hospitals infrastructure, patients or medical personnel.

Needless to say, for a serious health issue, a medical consult is paramount with no delay, and an electronic device cannot be able to substitute a detailed medical evaluation. In the case of the cancer or burned patients, that have undergone complex medical treatment such as chemical therapy and radiation therapy, and thus are inherently immunosuppressed or immunocompromised, it is utmost important to be isolated from the outside environment and

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contagious disease carrying vectors, represented by the medical personnel, medical equipment and patient relatives. Thus, special isolation rooms are required, a strict hygiene protocol and a minimum contact.

The MCT monitors and holters [5], usually use fewer electrodes attached to the body (3 to 5) and are more prone to electrical, electromagnetic and electrostatic noise, compared to the fixed medical grade ECG units [6], found in the ERs (emergency rooms) and ICUs (intensive care units), that use up to 12 electrodes for more precise data acquisition, while using ground shielding in the wires, for reducing the vast amounts of EMI (Electromagnetic Interference) noise from the other medical equipment, and high frequency wireless communication interference from mobile phones or proximity RF antennas.

2. THEORETICAL ASPECTS AND PROTOTYPE DEVELOPMENT

A common configuration of a MCT device with 3 sticky AgCl gel electrodes is described in Fig. 1. The *RA*--, *LA*+- and *LL*++ form the Einthoven's triangle [7], which is an imaginary formation for electrode lead placement points, for measuring voltage potential differences between them, of cardiac activity.

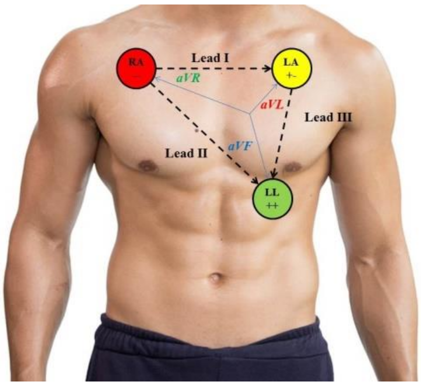

Figure. 1. Typical electrode positioning: RA-is right arm lead, LA-is left arm lead and LL-is left leg lead.

Theoretically, the unipolar augmented vector leads *aVR(-150º), aVL(-30º)* and *aVF(+90º)* are vectors derived from the respective electrodes *RA*, *LA* and *LL* in relation to the heart position.

In any ECG PQRST complex wave, PR and ST represents a combination of graphical deflections which corresponds to the depolarization and repolarization of the atria and ventricles inside the heart (Fig. 2) [8].

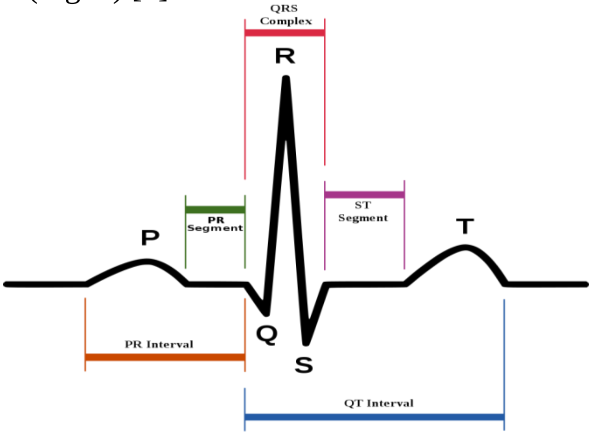

Figure. 2. Theoretical PQRST complex, PR segment and ST segment of a complete heart beat cycle.

The proposed mobile ECG open-source system is a low cost device, mainly comprised of an integrated signal conditioning block chip (AD8232) [9], soldered on a development adapter board. It is a low power module with standby function when no signal is detected, with 3.3V_{DC} input supply voltage and an optional shutdown pin to further cut the power consumption when no cardiac activity is detected (Fig. 3.).

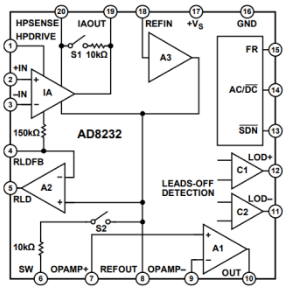

Figure. 3. AD8232 chip - functional block diagram.

The adapter board uses three outputs, one being the analog signal output *(OUT)*, and two being the digital "*leads off comparator output*" *(LOD+ and LOD-)* used for the detection mode when the electrodes are connected to, or disconnected from, the patient. Two similar boards were used for experimentation, one being from a branded developer and distributor, while the other being a clone from an unknown manufacturer (Fig. 4.).

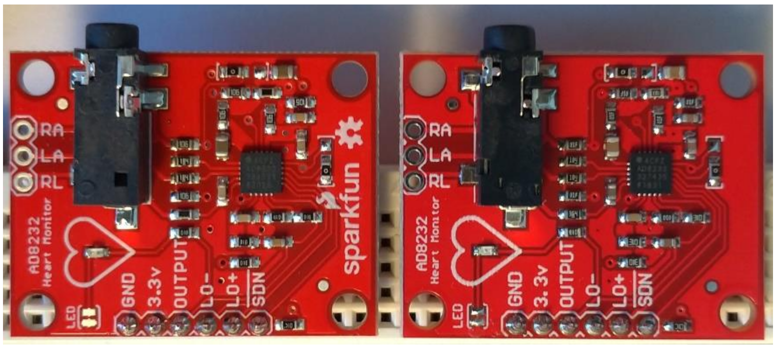

Figure. 4. AD8232 chip on an original adapter board (left) and cloned adapter board (right).

The two boards look quite similar, although some visual details and passive components values are different. In order to test them, a high performance programmable data acquisition board was used, such as ESP32, with wireless capabilities like WiFi and BLE, a dual 12 bit ADC (Analog to Digital Converter) and a dual core processor with a speed of 240MHz.

The proposed MCT system (Fig. 5) [10] is able to display on a Nokia 5110 GLCD screen, the real-time graphic values of the ECG captured from the patient's heart activity. The system uses 3 electrodes for fast data acquisition, with an approximate sampling rate of 1000 samples per second. Also in order to reduce the electrical background noise, a clamp-on ferrite core is attached to the three electrode wires.

The experimental device can either be powered by a conventional linear power supply that uses a LDO (Low-Dropout Regulator) in order to mantain a steady voltage with minimum losses, or by battery means, with a rechargeable Li-Po or Li-ion power-bank. Using a direct SMPS (Switch Mode Power Supply) would be a poor choice because of the switching frequency of the regulator, which can induce interference in the ADC of the ECG module, thus distorting the readings.

The best way of powering the device from the AC line is by using an opto-isolated, linear power supply to avoid the inherited 50-60Hz operation frequency of the AC line and electrical noise from local appliances or other EMI (Electromagnetic interference) generating sources [11].

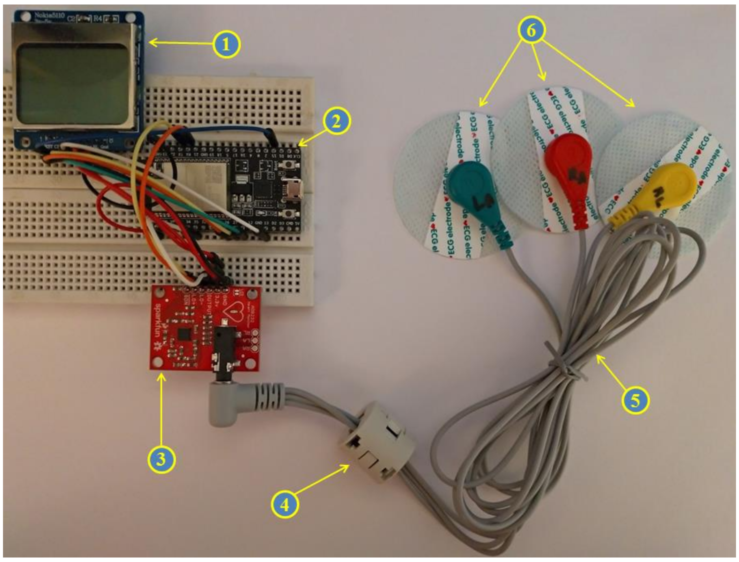

Figure. 5. Developed MCT system without the portable rechargeable battery as follows: 1-Nokia 5110 GLCD; 2-ESP32 development board; 3-AD8232 sensor board; 4-clamp-on ferrite core; 5- unshielded electrode wires; 6-electrodes with gel sticky pads.

The proposed algorithm built for reading the ECG values and heart rate (bpm) is described in Fig. 6 as a block diagram.

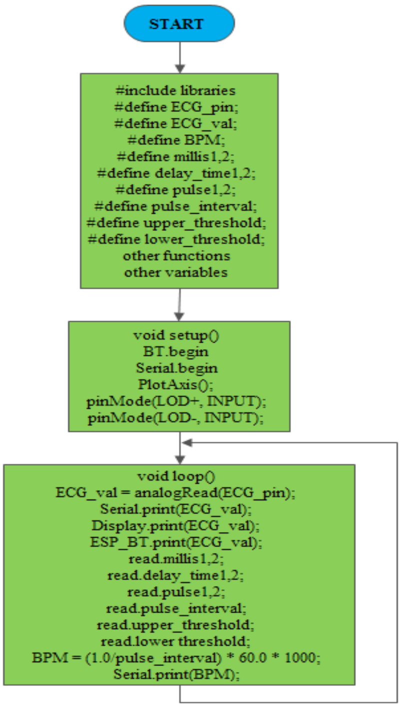

Figure. 6. Simplified algorithm flowchart for the ESP32 platform with ECG and bpm measurements.

While conducting measurements, the ESP32 platform is simultaneously sending the ECG analog values as streaming graphs to the GLCD, to the smartphone bluetooth graphic application and to the serial port. The bpm values are calculated according to the respective formula (1) and sent to the serial port aswell, for viewing. The portable MCT can also be connected to a PC using the USB port, in order to view the larger graphs on a LCD monitor using the serial graph function of the Arduino software.

On the other hand, the bluetooth graphic software on the smartphone can also store log files with the captured analog values of the ECG, for an indefinite time, thus enabling the user or the medic to have access to consistent ammounts of data about the cardiac activity events.

3. EXPERIMENTAL RESULTS AND CALCULUS

The experimental MCT measurements were conducted on a 40 year old male patient, weighing 80 kg, 1.79 m height, with no history of cardiac condition or other cardiovascular diseases.

The acquired data from the proposed MCT system during the measurements consisted in three ways:

- GLCD screen streaming graphics of maximum 2 beats per frame;
- Serial port streaming data – for ADC values, ECG and bpm;
- BLE wireless communication streaming graphics and data logging with smartphone BT application.

Also an important mention is that the system was connected to a PC usb port during the measurements (SMPS based power supply), thus a certain degree of inherited AC line and switching noise is present in the ECG data graphs.

Giving the small form factor, low power consumption and low resolution of the Nokia 5110 GLCD screen, its use, although limited, it can come to aid when no smartphone or PC is available to connect with the MCT system to have a broader view of the ECG graph. Fig. 7 describes a two pulse event capture of the ECG on the GLCD screen, during measurements.

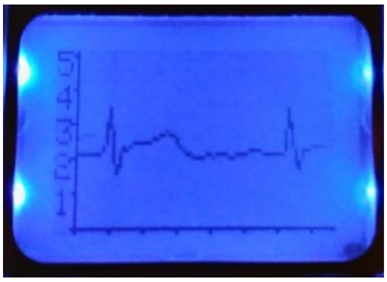

Figure. 7. GLCD screen ECG stream capture with branded adapter.

Using a serial port capture of the measurements, the PQRST complex is identified and highlighted in the ECG graph of the branded module (Fig. 8.).

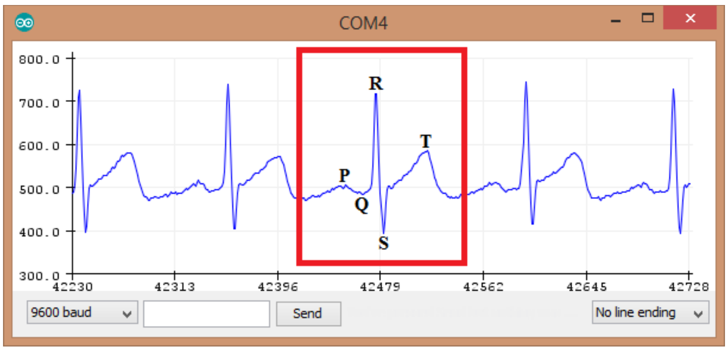

Figure. 8. Serial port graph.

During the experimentation process, with the two ECG modules, a consistent baseline drift and a small AC power line noise [12], of the ECG output signal, was observed on the clone adapter board described in Fig. 9.

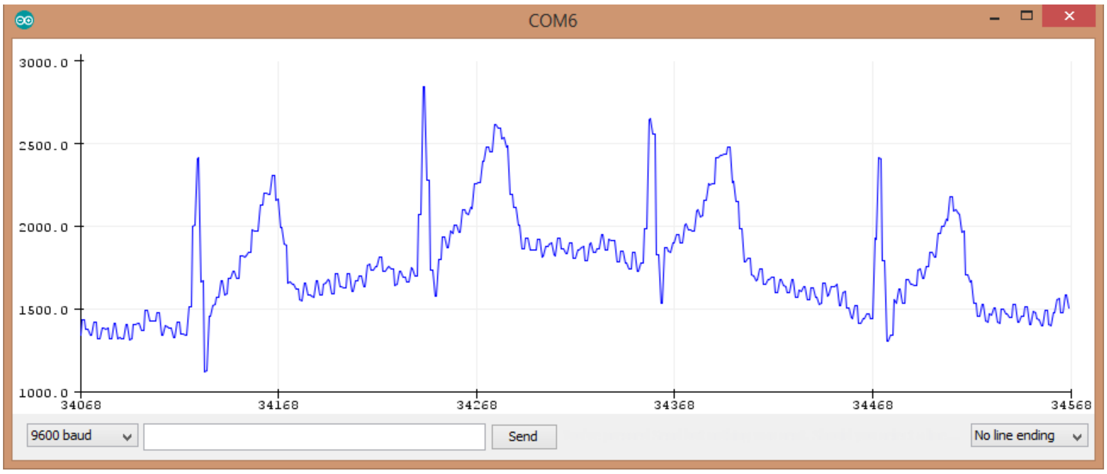

Figure. 9. AD8232 clone module ECG serial port graph capture.

On the other hand, the branded module ECG signal was more distinguishable and symmetrical regarding the amplitude and time period, thus being easier to read and interpret the PQRST diagram, as described in Fig. 10.

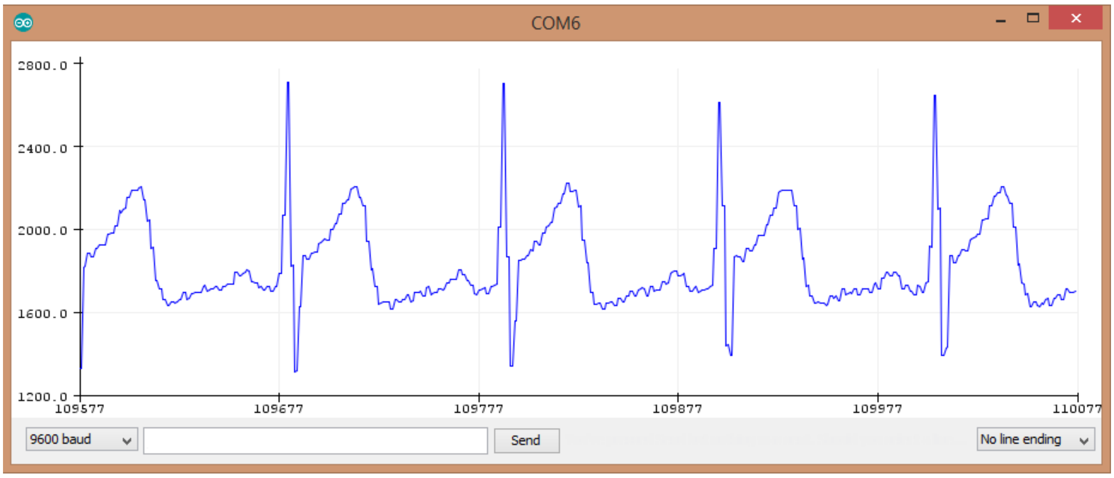

Figure. 10. AD8232 branded module ECG serial port graph capture.

Although the two modules are almost identical, the degree of the captured noise in the ECG measurements were quite different, a small conclusion is drawn here, that either the external passive filtering on the clone board was not in accordance to the AD8232 chip's manufacturer requirements, or the chip itself is counterfeited, thus lacking manufacturer standards.

The bluetooth tested range of the prototype MCT system, was about 30 meters inside a concrete building, without data loss. A screen capture with the ECG signal from the smartphone bluetooth application can be found in Fig. 11.

A major advantage of using bluetooth serial data streaming instead of a WiFi API based data streaming server, is the speed and the amount of the data points rendered in realtime, on the chart. While the PHP-HTML server protocol is limited by the request/response processes delays and limited bandwidth for the huge amounts of streaming data per second comming from the MCT, the simple bluetooth serial transmission protocol can easily overcome these issues.

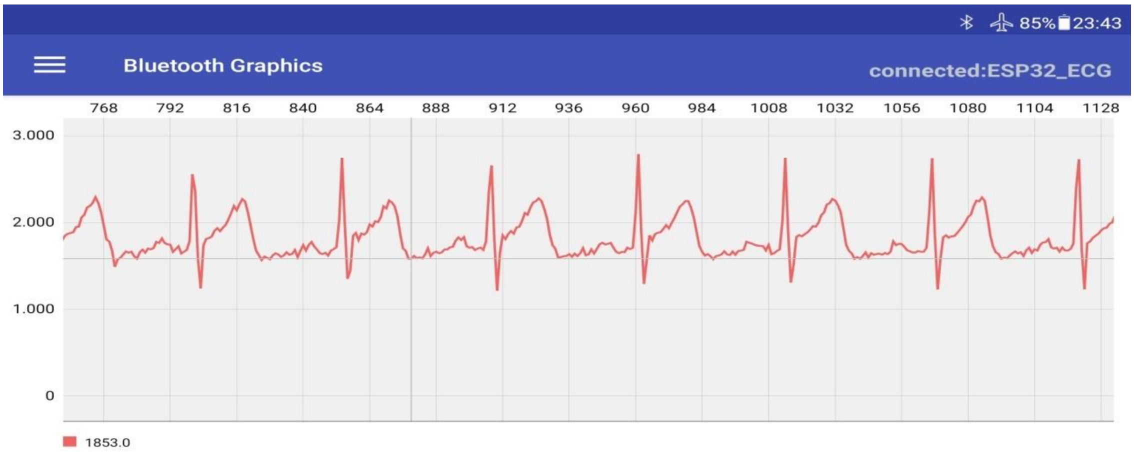

Figure. 11. Bluetooth smartphone ECG stream capture with branded adapter.

Another algorithm implementation in the MCT system was the heart rate capability (bpm), by calculating the pulse interval or time in ms between the first pulse and the second pulse, by using the frequency relation (1).

$$HR_{[bpm]} = \frac{1}{T_{[ms]}} \times 10^{3} \times 60_{[s]} \tag{1}$$

where $HR_{[bpm]}$ is the heart rate in *beats per minute*, $T_{[ms]}$ is the period interval in *milliseconds* and $10^{3} \times 60_{[s]}$ is the conversion factor to minute.

In Fig. 12 a serial port capture is made for the reading of each pulse value that contains R and S peaks with the subsequent bpm. Thus, knowing that the R of one ECG cycle is the highest value pulse point and S is the lowest value pulse point, in relation with a fixed upper threshold amplitude value mark (~3200) and a fixed lower threshold amplitude value mark (~1200), correlated with the pulse interval, the heart rate could be calculated and displayed.

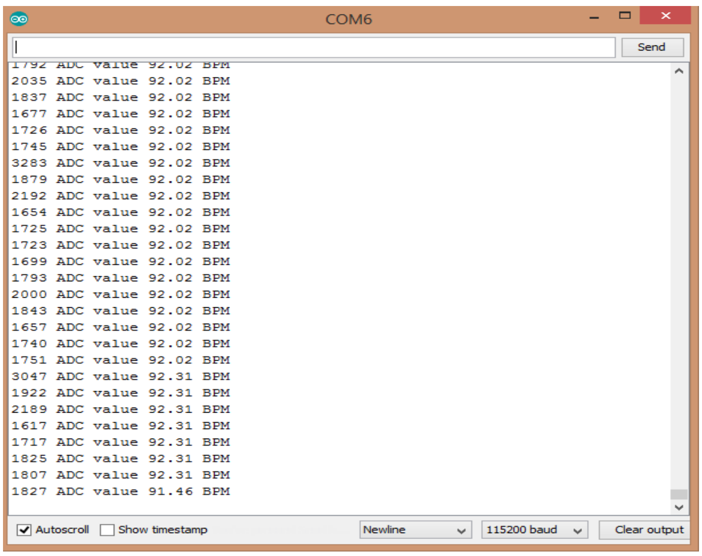

Figure. 12. ECG serial port capture of ADC values and heart rate with the branded module.

The ADC signal numeric values of the ECG can be calculated from relation (2), for each acquired sample:

$$ADC_{val} = \frac{V_{in} \times ADC_{res}}{V_{sys}} = \frac{V_{in} \times 2^{12}}{3.3V_{DC}} \tag{2}$$

where ADC_{val} is the read value from the ECG sensor, V_{in} is the input signal voltage, ADC_{res} is the resolution of the ESP32 ADC module of 4095 bits and V_{sys} is the ESP32 system operating voltage of $3.3V_{DC}$.

According to the data points extracted from the data logs, the MCT system makes 60 sample reads for a complete heart beat and comparing the experimental bpm value series of the MCT system, versus a commercial pulse oximeter, the error was about 1% during measurements.

An estimative current consumption calculus (3) is used, in order to choose the optimal power bank *mAh* capacity, for maximum autonomy and small form factor, of the experimental MCT system.

$$I_{tot} = \sum_{k=1}^{N} i_{nk} = i_{ESP32} + i_{GLCD} + i_{ECG} = 130 + 20 + 10 = 160[mA] \tag{3}$$

where I_{tot} is the total current consumption of the MCT system in *milliamperes*, N is the total number of consumers, i_{nk} is the current consumption of each device, i_{ESP32} is the current consumption of the ESP32 platform with the Rx/Tx (Receive/Transmit) BLE module active, i_{GLDC} is the current consumption of the Nokia 5110 LCD, i_{ECG} is the current consumption of the ECG module board.

Just to mention, the measured current draw from the external Li-ion power bank is made at $5V_{DC}$ on the USB power input of the ESP32 development board, by the means of an USB digital power meter (Fig. 13).

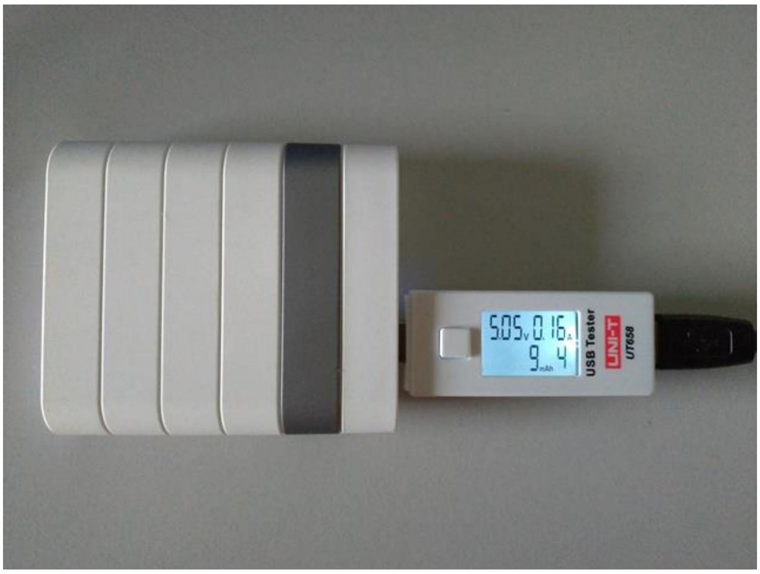

Figure. 13. The experimental MCT system connected to a 10200mAh power bank, through an USB digital power meter – reading a constant voltage of 5.05V$_{DC}$ and a 0.16A current draw, while conducting the ECG measurements.

Corroborating the I_{tot} results with the required power bank capacity, relation (4) is obtained:

$$C_{bat} = [A] \times [h] \times 1000 = I_{tot} \times h \times 1000 = 0.16 \times 64 \times 10^3 = 10240[mAh] \quad (4)$$

where C_{bat} is the required power bank capacity in *mAh*, *A* is the current consumption and *h* is the number of operation hours. As a result, a 10200 mAh power bank is suited for the proposed MCT that can roughly last for two and a half days of full operation.

4. CONCLUSIONS

In this paper, the research was conducted regarding a proposed MCT system, with low costs, versatile, accurate, low power consumption and thus highly autonomous, which revealed that although it can perform almost as good as its commercial counterparts, more research, study and testing is still required in order to represent a finished competitive solution.

In conclusion, there is still a lot of room for improvements, and thus, future developments of this prototype would include functions such as: shielded cables for the electrode probes, better external passive filtering for AC noise suppression, battery power consumption optimisation with MCT sleep/wake trigger events, WiFi or GPRS event data transmission with IoT support, enhanced LCD graphics, heart abnormality detection, improved bpm reading algorithm and additional software filtering.

A final thought would be that this device cannot represent a substitution for a medical diagnostic, but it can represent a good health assistant in critical situations or remote scenarios.

REFERENCES

[1] Fye, W.B., *The American Journal of Cardiology*, **73**, 937, 1994.
[2] Vasile, I., Vasile, V., Diaconu, E., Andrei, H., Angelescu, N., *Journal of Science and Arts*, **48**, 793, 2019.
[3] Raka, A.G., Naik, G.R., Chai, R., *Applied Sciences*, **7**, 954, 2017.
[4] Tsang, J. P., Mohan, S., *Medical devices (Auckland, New Zeeland)*, **7**, 1, 2013.
[5] Kułach, A., Dewerenda, M., Majewski, M., Lasek-Bal, A., Gąsior, Z., O*pen medicine (Warsaw, Poland)*, **15**, 697, 2020.
[6] Xu, Y., Luo, M., Li, T., Song, G., *Sensors (Basel, Switzerland)*, **17**, 2754, 2017.
[7] Yoneyama, K., Naka, M., Harada, T., Akashi, Y., *Journal of Arrhythmia*, **36**, 1107, 2020.
[8] Cano, M. E., Jaso, R. A., Tavares, M. E., Estrada, J. C., Mena, E. A., Reynoso, O., González Vega, A., Córdova Fraga, T., *Mexican Journal of Biomedical Engineering*, **32**, 9, 2018.
[9] Kubov, V. I., Dymytrov,Y. Y., Stojanović, R., Kubova, R. M., Škraba, A., *Proceedings of the 9th Mediterranean Conference on Embedded Computing (MECO)*, Montenegro, 1, 2020.
[10] Marouf, M., Vukomanovic, G., Saranovac, L., Bozic, M., *Biomedical engineering online*, **16**, 80, 2017.

[11] González-Briceño, G., Medow, J. E., Ortega-Cisneros, S., Lorias-Espinoza, D., Minor-Martínez, A., Webster, J. G., *IEEE Transactions on Instrumentation and Measurement*, **68**, 797, 2019.
[12] Akwei-Sekyere S., *PeerJ*, **3**, e1086, 2015.