XIX IMEKO World Congress
Fundamental and Applied Metrology
September 6−11, 2009, Lisbon, Portugal

# A MULTICHANNEL WIRELESS EMG MEASUREMENT SYSTEM BASED ON INTRABODY COMMUNICATION

Željka Lučev [1], Igor Krois [2], Mario Cifrek [3]

[1, 2, 3] University of Zagreb, Faculty of Electrical Engineering and Computing, Zagreb, Croatia
[1] zeljka.lucev@fer.hr, [2] igor.krois@fer.hr, [3] mario.cifrek@fer.hr

**Abstract** − In this paper we present the novel implementation of the intrabody communication (IBC) system that was specially designed for electromyography (EMG) measurements in kinesiology, sports medicine and rehabilitation. We propose a novel approach to the wireless EMG monitoring system design, which prolongs the battery life by minimizing the power consumption requirements for data transmission. This goal was achieved by a capacitive IBC approach and by developing special-purpose ultra-low power hardware modules, which perform tasks of digital signal modulation and demodulation at a very low-level. We investigate the optimal electrodes placement for IBC and present the results of *in vivo* measurements.

**Keywords**: intrabody communication, wireless EMG system, electrode arrangement

## 1. INTRODUCTION

Intrabody communication (IBC) is a relatively new type of communication which uses the human body as a signal transmission medium. It was first described in 1995 by Zimmerman [1], who used it for communication between electronic devices on and near the human body by capacitively coupling picoampere currents through the body. Two main methods of intrabody signal transmission are capacitive and galvanic, which differ by the basic principle of the signal coupling. In the capacitive coupling approach [2-4] the induced electrical signal is controlled by an electric potential, while in the galvanic coupling solution [5, 6] the communication signal is controlled by an alternating current flowing through the human body. In the capacitive coupling approach, the signal path closes through the human body between the transmitter unit signal electrode (TX S) and the receiver unit signal electrode (RX S), while the return signal path is established between the transmitter unit ground electrode (TX G) and the receiver unit ground electrode (RX G). In such a setting, the signal transmission path highly depends on the surrounding environment. One of the main problems in establishing the intrabody communication system is how to arrange the signal and ground electrodes properly on the body surface in order to obtain the maximum signal strength at the receiver unit.

Intrabody signal communication is a particularly interesting approach for data transmission which can be applied in the field of electromyography (EMG) measurements and monitoring in kinesiology and sports medicine. Traditionally employed wired EMG electrodes have numerous shortcomings in biomechanical monitoring, such as complex installation of the measurement system, disturbance of the normal course of the exercise, and easy detachment of electrodes (due to rapid body movements, e.g. in athlete monitoring), thus deteriorating the results of the measurement. Therefore, there is a need for wireless and lightweight active electrodes which can be placed on the skin surface more simply and that will not affect usual athletic movements.

There are few commercially available battery powered products that employ the idea [7, 8] of the wireless EMG system consisting of few transmitters and a single central receiver. However, they use 2,4 GHz (WiFi) radio-frequency data-link and have rather high power requirements, what is further aggravated by the fact that the receiver unit is situated few meters away from the body. The continuous EMG signal transmission using the radio-frequency link can operate only a few hours without replacing the batteries, hence affecting the practical system usability in long-term monitoring applications.

In this paper we propose a novel approach to the wireless EMG monitoring system design, which prolongs the battery life by minimizing the power consumption requirements for data transmission. This goal was achieved by a capacitive IBC approach and by developing special-purpose ultra-low power hardware modules, which perform tasks of digital signal modulation and demodulation at a very low-level, thus providing the means for flexible control of the overall system power consumption. The receiver unit is placed on the body at the short distance from the transmitters, hence further reducing the power consumption. We investigate the optimal electrodes placement for IBC and present the results of *in vivo* measurements.

The paper is organized as follows. In Section 2 we propose the novel wireless EMG system design concept. We discuss the choice of the carrier frequency and present the transmitter and receiver unit implementation. We describe the test measurement setup in subsection 3.1 and investigate propagation through the air and body in subsection 3.2. Also, we present the results of comparison between several electrode placements in subsection 3.3. Final conclusions and remarks are provided in Section 4.

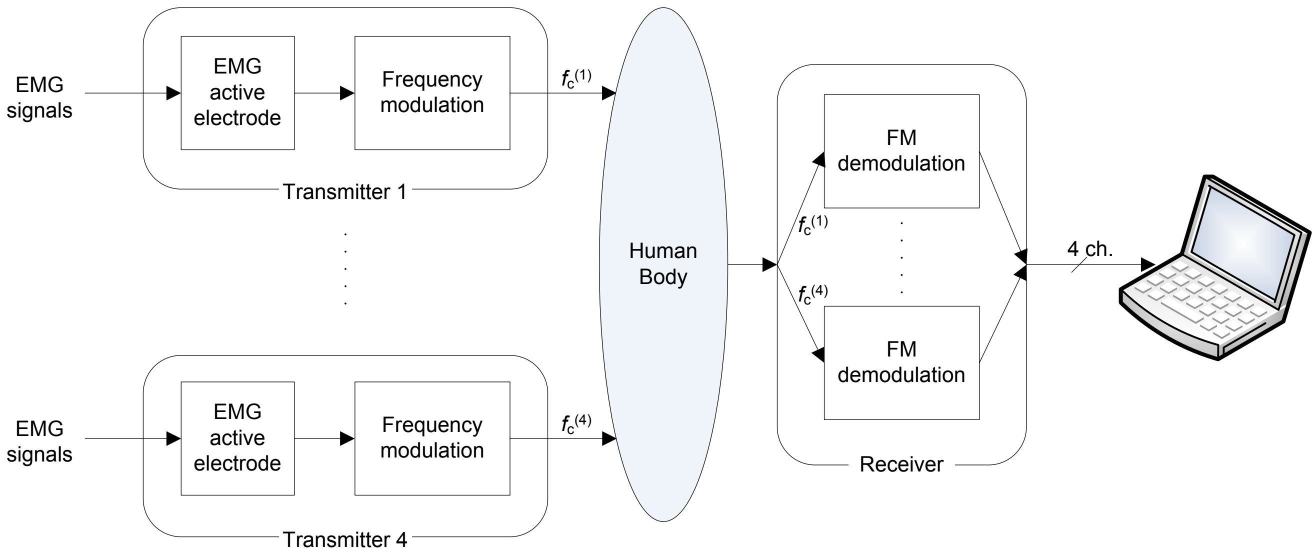


Fig. 1. EMG intrabody communication system.

## 2. SYSTEM DESIGN

The IBC hardware consists of two main parts: four transmitter modules and the four-channel receiver unit, connected as shown in Fig. 1. The frequencies $f_c^{(1)}$–$f_c^{(4)}$ of four signal carriers for intrabody communication are close to 1 MHz.

### *2.1. Carrier frequency*

The selection of the signal carrier frequency is a trade-off between several requirements:

- constraints imposed by the safety regulations,
- low-frequency communication data link (due to the ultra-low power requirements),
- high tissue conductance at the signal carrier frequency (achieved at higher frequencies).

Safety regulations for limiting the exposure to the time-varying electric, magnetic and electromagnetic fields are based on the international guidelines provided by the International Commission on Non-Ionizing Radiation Protection (ICNIRP) [9]. In the frequency range of 100 kHz–10 GHz the recommended whole-body average SAR (Specific Absorption Rate) is 0,08 W/kg, with a maximum allowed contact current of 20 mA. Moreover, the signal transmitted through the human body must not interfere with normal biological signals, meaning that its frequency should be higher than 1000 Hz.

Fig. 2 depicts the frequency dependence of conductivities of the human tissues relevant for the intrabody signal transmission (skin, muscle, fat and bone) [10]. The signal propagates best through the skin and muscles due to their high conductivity, Fig 2.

At the frequency of 1 MHz conductivities of wet and dry skin differ less than at the frequency of 100 kHz. Since the system is supposed to be worn by athletes during physical activity, it is more likely that their skin will be wet, therefore the value of the skin conductivity will be closer to that of the wet skin. At the frequencies between 1 MHz and 1 GHz the muscle and the wet skin have high and almost constant conductivity and therefore, considering the low power requirements, it is better to use the signal of the lower frequency (1 MHz). Above 100 MHz the conductivity of all tissues increases rapidly, as well as the power consumption.

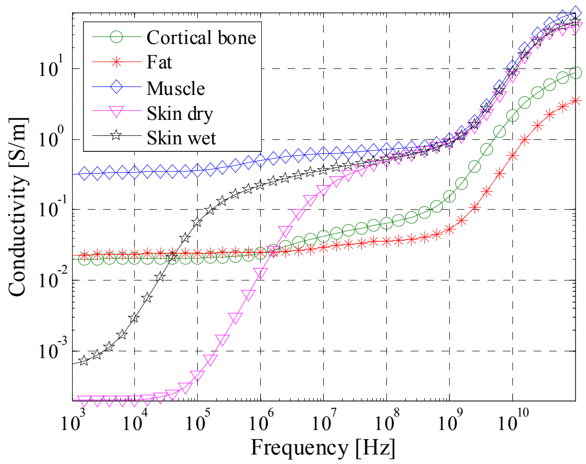


Fig. 2. Frequency dependence of conductivities for different human body tissues. Adapted from [10].

### *2.2. Transmitter unit*

The transmitter input stage consists of an active EMG electrode with high input impedance that registers biological signal whose frequency is between 15 Hz and 500 Hz. The amplified EMG signal is fed to the 12-bit A/D converter and adjusted to match the format of the frequency word of the DDS integrated circuit. The DDS circuit is used for analog frequency modulation of the signal (FM), which is controlled by a PIC12C509 microcontroller. The output of the TX signal electrode is a frequency modulated sine wave with an amplitude of 1,15 $V_{pp}$, approximately 1 MHz carrier frequency, 5 kHz frequency deviation and is well within the limits of the safety regulations [9]. Dimensions of the transmitter unit are 4,5 cm x 2 cm x 1 cm. The module is powered by a single 1,5 V battery.

Four identical transmitter units communicate with the receiver independently using the frequency domain multiplex. Each module uses a slightly different carrier frequency ($f_c^{(1)}$ = 0,999 MHz, $f_c^{(2)}$ = 1,047 MHz, $f_c^{(3)}$ = 1,0941 MHz, and $f_c^{(4)}$ = 1,141 MHz) and 5 kHz frequency deviation.

### *2.3. Receiver unit*

The receiver unit of the wireless EMG system consists of four identically designed channels, with an exception of the first channel which contains an extra reference oscillator. Each channel incorporates a heterodyne receiver, which is used for FM signal demodulation. The heterodyne receiver was implemented using SA608 integrated circuit. Although this integrated circuit was originally intended for use in a high-frequency range (150 MHz), we achieved a good performance in 1 MHz range by few modifications of the original application circuit implementation. Dimensions of the receiver unit are 10,5 cm x 6 cm x 2,5 cm. The maximum current consumption from 3 V battery supply is 26,5 mA, when all four channels operate simultaneously. The logarithmic received signal strength can easily be obtained using the RSSI pin on each channel of the receiver.

## 3. RESULTS AND DISCUSSION

### *3.1. Measurement setup*

For connecting transmitter and receiver units to the human body we used BlueSensor Ag/AgCl electrodes. The skin was prepared by using an abrasive gel. The receiver was connected to the data acquisition card attached to the battery powered personal computer, in order to avoid ground loops that would deteriorate our measurements. For easier demonstration of the hardware's functionality with the human body as a transmission path, the signal being modulated is not an EMG signal. Instead, the transmitter's microcontroller generates the signal with the following characteristics: saw waveform, frequency 101,65 Hz, amplitude 0,145 $V_{pp}$. All measurements are made using the transmitter with 1,047 MHz carrier frequency.

### *3.2. Propagation through the air*

Firstly, we needed to verify that the signal transmission path was established through the subject's body and not through the environment (air). Signal electrodes of the transmitter and the receiver were attached to the subject's left arm, ground electrodes remained unconnected and the distance between the signal electrodes varied along the arm. The whole procedure was repeated while changing the distance between the signal electrodes in the air.

Fig. 3 shows received signal strengths acquired from the RSSI pin of the corresponding receiver channel. The measured values are between -54 dBm and -62 dBm for intrabody signal propagation and approximately -99 dBm for propagation through the air, what is equal to the noise level. Fig. 4 presents demodulated waveforms at the distance $d$ = 31 cm for both propagation paths.

Fig. 3 and Fig. 4 show that the intrabody propagation is superior to the propagation through the air at the 1 MHz frequency for distances up to 31 cm. It is reasonable to assume that the same relationship is valid for greater distances as well.

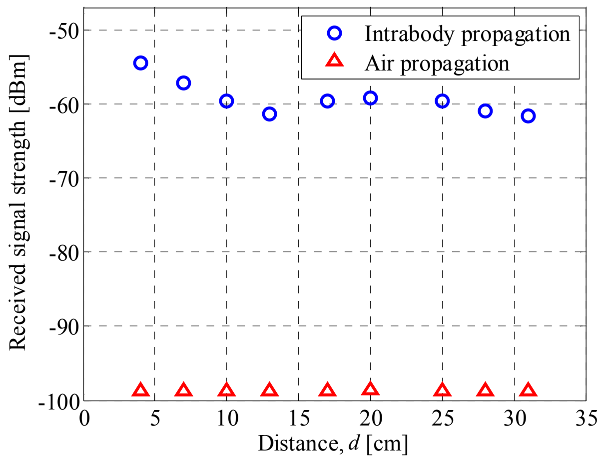


Fig. 3. Comparison of intrabody and air propagation.

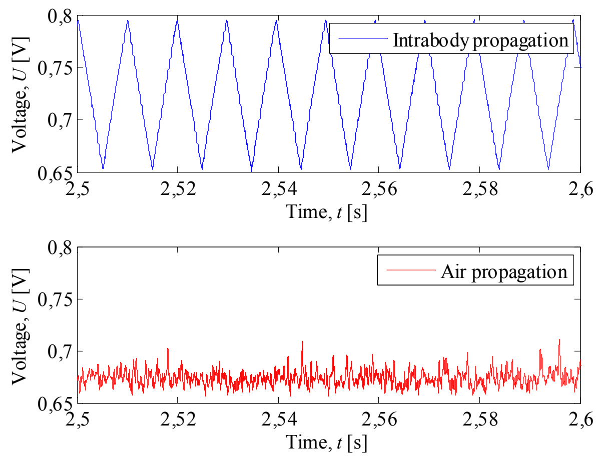


Fig. 4. Received signals for $d$ = 31 cm.

### *3.3. Comparison of electrode placements*

The received signal level is highly affected by the orientation of the transmitter with respect to the receiver and the number of ground electrodes connected to the body.

According to [3, 11] a transmitter ground electrode close to the transmitter signal electrode strengthens the generated electric field, while the presence of a receiver ground electrode close to the receiver signal electrode reduces the received signal level. Therefore, the authors recommend that the signal and ground electrode of the transmitter and a signal electrode of the receiver should be in contact with the body, while the ground electrode of the receiver should remain disconnected. Moreover, the received signal strength is substantially higher for longitudinal direction of TX with respect to RX (position 2, Fig. 5) than for the transversal direction (position 1, Fig. 5) [3]. In both cases, the ground electrode of the receiver is disconnected from the body.

Unlike [3, 11], in [12] authors conclude that the optimal electrode arrangement is such that both signal electrodes are connected to the body and both ground electrodes remain unconnected. This electrode position is called a two-electrode arrangement and is shown as position 3 in Fig. 5.

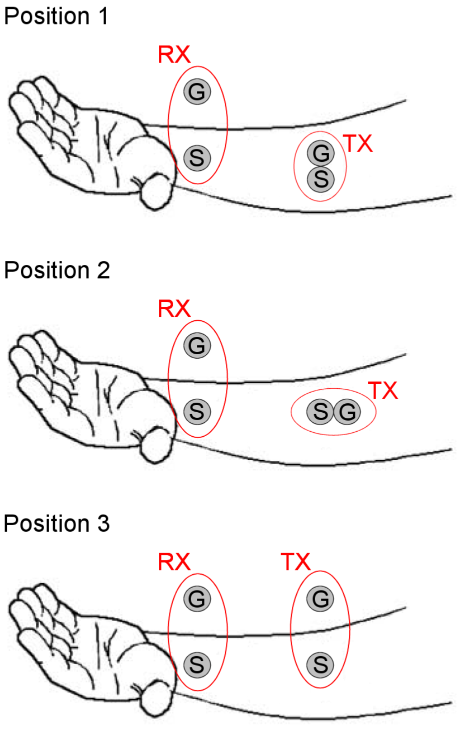


Fig. 5. Positions of the electrodes: position 1 – transversal direction, position 2 – longitudinal direction, position 3 – two-electrode arrangement. TX is a transmitter, RX is a receiver; S denotes a signal electrode and G denotes a ground electrode.

Since there is a certain ambiguity in literature about the optimal arrangement of electrodes in Fig. 5 that yields the largest received signal level, and since we applied lower excitation frequency (1,047 MHz instead of 10 MHz and 10,7 MHz as in [3, 11] and [12], respectively), we decided to evaluate all three proposed electrode arrangements. The summary of our measurements is depicted in Fig. 6.

The distance *d* assigned to the x-axis in Fig. 6 refers to the distance between the centre of the receiver and transmitter signal electrode (S RX and S TX). The receiver signal electrode S RX was fixed to the left wrist of the subject and the ground electrode G RX remained in the air. The transmitter was moved along the left arm from the wrist to the shoulder. The space between signal and ground electrodes of the transmitter in positions 1 and 2 was fixed to 4 cm. The distance *d* between the S RX and the transmitter varied from 4 cm to 31 cm. The position of the elbow was at the distance of 20 cm.

For the position 1 the received signal strength was the lowest for all three considered configurations and the signal transmission was the worst. For distances shorter than 10 cm, position 2 exhibits higher received signal strength and for distances longer than 10 cm, positions 2 and 3 show similar characteristics.

It is evident that the elbow position has influence on positions 1 and 3 and that it has no influence on position 2. The cause of this phenomenon is the structure of the tissue under the electrodes: there is no muscle tissue at the elbow and, as it has already been shown in Fig. 2, the conductivity of the muscle tissue is the highest of all the tissues relevant for the signal transmission. When *d* is 20 cm and the electrodes are arranged as in position 2, the ground RX electrode is positioned above the muscle tissue of the upper arm and the signal is therefore strengthened.

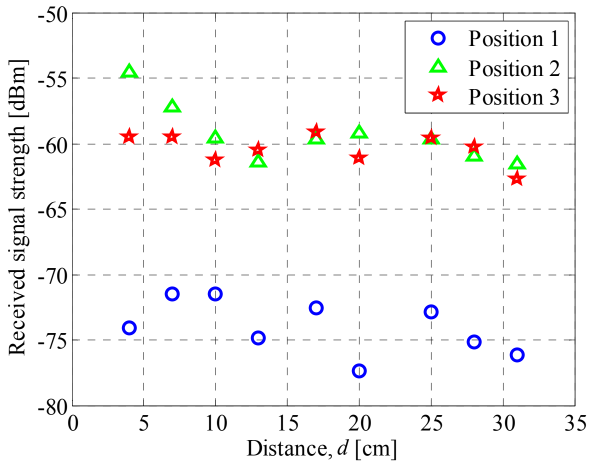


Fig. 6. Received signal strength for three different electrode positions along the left arm
(0 cm – wrist, 20 cm – elbow, 45 cm – shoulder).

## 4. CONCLUSIONS

We presented the novel implementation of the IBC system that was specially designed for electromyography measurements in kinesiology, sports medicine and rehabilitation. We showed that the signal transmission between our transmitter and the receiver through the human body is superior to the propagation through the air and we experimentally determined the optimal arrangement of the electrodes.

Currently we focus our attention to understanding differences between positions 2 and 3 for greater distances between electrodes and influence of the joints to the transmission characteristics. We will make measurements on more test subjects in order to statistically confirm the presented results. Future work will include further minimization of the power consumption as well as improvements in the mobility by implementing a wireless communication between the receiver and data acquisition unit.

## ACKNOWLEDGMENTS


This research has been supported by the Croatian Ministry of Science, Education and Sport through the research project “Noninvasive measurements and procedures in biomedicine”, grant number 036-0362979-1554.

## REFERENCES


[1] T. G. Zimmerman, "Personal Area Networks: Near-field intra-body communication", M.S. thesis, MIT Media Laboratory, Cambridge, MA, Sept. 1995.

[2] K. Partridge et al., "Empirical measurements of intrabody communication performance under varied physical configurations", *UIST '01: Proceedings of the 14th annual ACM symposium on User interface software and technology*, pp. 183–190, Orlando, Florida, 2001.

[3] K. Fujii et al., "Study on the transmission mechanism for wearable devices using the human body as a transmission channel", *IEICE Trans. Commun.*, vol. E88-B, nº. 6, pp. 2401–2410, June 2005.

[4] M. Shinagawa, M. Fukumoto, K. Ochiai and H. Kyuragi, "A near-field-sensing transceiver for intrabody communication based on the electrooptic effect", *IEEE Transaction on Instrumentation and Measurement*, vol. 53, pp. 1533–1538, Dec. 2004.

[5] T. Handa, S. Shoji, S. Ike, S. Takeda and T. Sekiguchi, "A very low-power consumption wireless ECG monitoring system using body as a signal transmission medium", *in Proc. of the 1997 International Conference on Solid-State Sensors and Actuators*, pp.1003–1006, Chicago, USA, June 1997.

[6] M. S. Wegmüller, M. Oberle, N. Kuster and W. Fichtner, "From dielectrical properties of human tissue to intra-body communications", *in Proc. of IFMBE, World Congress on Medical Physics and Biomedical Engineering 2006*, vol. 14, nº. 2, pp. 613–617, Seoul, Korea, Aug.–Sept. 2006

[7] Aurion ZeroWire EMG web site, (January 2009.) [Online], http://www.aurion.it/technical specification.pdf

[8] BTS FreeEMG web site, (January 2009.) [Online], http://www.bts.it/proser/freeemg.htm

[9] ICNIRP, "Guidelines for Limiting Exposure to Time-Varying Electric, Magnetic, and Electromagnetic Fields (up to 300 GHz)", *Health Physics*, vol. 74, nº. 4, pp. 494–522, April 1998.

[10] Dielectric properties of body tissues calculator, IFAC, (January 2009.) [Online], http://niremf.ifac.cnr.it/tissprop/

[11] K. Fujii, K. Ito and S. Tajima, "A study on the receiving signal level in relation with the location of electrodes for wearable devices using human body as a transmission channel", *IEEE Antennas and Propagation Society International Symposium*, vol. 3, pp. 1071–1074, June 2003.

[12] K. Hachisuka et al., "Simplified circuit modeling and fabrication of intrabody communication devices", *Sensors and actuators A: physical*, vol. 130–131, pp. 322–330, 2006.