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# Engineering Power

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## EDITOR-IN-CHIEF'S WORD

Dear readers,

in this issue of our international bulletin, whose guest editor is a distinguished member of our Academy, Prof. Mario Cifrek, PhD, from the Faculty of Electrical Engineering and Computing of the University of Zagreb, we have a special opportunity to learn about cooperation between scientists from China and Croatia, this time in the field of intrabody communication.

I believe that it will be interesting for you to get acquainted with this international research, especially because of the actuality of the subject. It will also be possible to make contacts and discuss with the authors.

Editor
Vladimir Andročec, President of the Croatian Academy of Engineering

## EDITOR'S WORD

Dear readers,

Although international collaboration has always been an important element of successful research, contemporary scientific activities are – almost 'by definition' – marked by joint efforts and a common research vision of the partners involved.

To this end, it is our pleasure to present in this edition of Engineering Power a fruitful research collaboration between the University of Zagreb, Faculty of Electrical Engineering and Computing and the Fuzhou University, China.

The Guest-Editor is Mario Cifrek, Member of the Academy and Professor at the Faculty of Electrical Engineering and Computing, University of Zagreb.

Editor
Zdravko Terze, Vice-President of the Croatian Academy of Engineering

## FOREWORD

**Ten years of collaboration between the University of Zagreb, Faculty of Electrical Engineering and Computing and Fuzhou University**

It has been 24 years since the entry into force of the "Agreement on scientific and technological cooperation between the Government of the Republic of Croatia and the Government of the People's Republic of China". In that period, a total of 165 scientific research projects have been approved. Our story begins ten years ago, in 2011, when a young assistant professor at Fuzhou University, Yueming Gao, contacted a doctoral student at the University of Zagreb, Faculty of Electrical Engineering and Computing, Željka Lučev, with an initiative to apply for a bilateral project under that Agreement. The backbone of the research was a topic they both worked on: intrabody communication (IBC), a wireless communication technology that uses living tissues as a transmission medium.

Both research groups already had noticeable experience in the field at the time. Shortly after the publication of Thomas G. Zimmerman's master's degree thesis, "Personal Area Networks (PAN): Near-Field Intra-Body Communication" in 1995, Prof. Igor Krois and Prof. Mario Cifrek initiated research at the Faculty of Electrical Engineering and Computing on the topic of capacitive intrabody communication. Colleague Željka Lučev joined the group in 2007 as a doctoral student. At the same time, colleague Yueming Gao from Fuzhou University, Key Laboratory of Medical Instrumentation & Pharmaceutical Technology, worked on intrabody communication based on galvanic coupling. His mentors were Prof. Min Du from Fuzhou University and Prof. Mang I Vai from Macau University. The bilateral project launched in 2011 opened the opportunity to exchange knowledge and expertise that both groups have acquired about different modalities of communication using the human body. In recent years, research has been expanded to the electrical impedance myography and monitoring of physiological parameters. Over the past ten years, seven researchers from Croatia and five from China - have participated in the research, while four doctoral theses and 37 master's degree theses have been written on these topics. The collaboration has so far resulted in a total of seven projects, eleven journal papers, and sixteen conference papers.

The introductory part of this issue chronologically presents the development of cooperation, mutual visits of HR and KIN researchers and delegations, joint projects, and a complete bibliography of both groups. Thanks to our Chinese partners, this part of the text has also been translated into Chinese. The following are two review articles summarizing research conducted over the past ten years. The first article summarizes the results of the research in the field of intrabody communication, a topic on which the cooperation was initially initiated. The following is an article describing research launched in 2018 on a new topic, electrical impedance myography for muscle fatigue monitoring. The ultimate goal of this research is the development of a small size wearable device for muscle fatigue monitoring.

Guest-Editor
Mario Cifrek, PhD, University of Zagreb, Faculty of Electrical Engineering and Computing

## CONTENT



*Željka Lučev Vasić[1], Yueming Gao[2], Min Du[1], Mario Cifrek[1]*

# TEN YEARS OF COLLABORATION BETWEEN UNIVERSITY OF ZAGREB, FACULTY OF ELECTRICAL ENGINEERING AND COMPUTING AND FUZHOU UNIVERSITY

[1] University of Zagreb, Faculty of Electrical Engineering and Computing
[2] Fuzhou University, College of Physics and Information Engineering

Back in February 2011, a young assistant professor Yueming Gao from Fuzhou University, College of Physics and Information Engineering contacted a Ph.D. student Željka Lučev Vasić from University of Zagreb, Faculty of Electrical Engineering and Computing with a suggestion to prepare a common project proposal for the Croatian-Chinese Scientific and Technological Cooperation call. At the time, both groups were working on different aspects of intrabody communication (IBC) topic, exploring how the human body could become a part of communication channel between electrical devices in its vicinity. The project proposal „Research of Intrabody Communication for Body Area Networks“ was prepared and submitted in April, and funding was approved in October 2011. Principal investigators on this project were Prof. Mario Cifrek from the University of Zagreb (FER) and Prof. Mang I Vai from Fuzhou University, College of Physics and Information Engineering (FZU) and University of Macau, Faculty of Science and Technology (UMAC). Researchers from all three institutions were involved in the project, Fig. 1.

**Figure 1.** Collaborative universities: Fuzhou University, University of Zagreb, University of Macau.

The collaboration between Croatian and Chinese researchers has been going on ever since, in the field of intrabody communication and wireless devices and networks for health and rehabilitation monitoring. Main research topics were galvanic and capacitive IBC system modeling, IBC channel measurements and IBC hardware design, which were recently expanded with electrical impedance myography and monitoring of physiological parameters.

Croatian and Chinese researchers have similar academic backgrounds and research fields, but are focused on different aspects of similar research questions. The Croatian side (FER) has a rich experience in the field of capacitive intrabody communication, muscle fatigue evaluation in biomechanics, biomedical sensors (electromyography, EMG; electrocardiography, ECG; electroencephalography, EEG), ultra-wideband (UWB) and biomedical signal processing. The Chinese side (FZU and UMAC) is good at galvanic intrabody communication, electromagnetic modeling of biomedical systems, real-time detection of diverse biochemical parameters, and the design of medical devices. They also have experience in the industrialization and certification of developed medical devices.

Until now, the collaboration with Prof. Gao and his team resulted in four bilateral projects and two projects funded by Fujian province:

- 2011 – 2013 "Research of Intrabody Communication for Body Area Networks," bilateral project, PI Prof. Mario Cifrek and Prof. Mang I Vai;

- 2013 – 2015 “Cooperative Study of the Multi-Coupling Type Intra-body Communication for Body Area Networks,” International cooperation project of MOST of China, PI Prof. Mang I Vai and Prof. Mario Cifrek;
- 2015 – 2017 “Intrabody Communication as a Key Technology for Internet of Things in Health Applications,” bilateral project, PI Prof. Mario Cifrek and Prof. Yueming Gao;
- 2018 – 2020 “Body area networks for health applications based on intrabody communication,” bilateral project, PI Assis. Prof. Željka Lučev Vasić and Prof. Yueming Gao;
- 2018 – 2021 “Evaluation of the Local Muscle Fatigue with the EIM Method for Wearable Applications,” project of S&T Department of Fujian Province, China, PI Assis. Prof. Željka Lučev Vasić and Prof. Yueming Gao;
- 2020 – 2022 “Body Area Network for Athlete Fatigue Monitoring,” bilateral project, PI Prof. Mario Cifrek and Prof. Yueming Gao;
- 2021 – 2024 “Real-Time Impedance Spectroscopy of Low Back Muscles Based on Multi-Frequency Excitation,” project of S&T Department of Fujian Province, China, PI Assoc. Prof. Željka Lučev Vasić and Prof. Yueming Gao.

The Croatian researchers involved in these projects were Assist. Prof. Željka Lučev Vasić, Prof. Mario Cifrek, Prof. Igor Krois, Prof. Silvio Hrabar, Assist. Prof. Josip Lončar, Ivana Čuljak, M.Sc., and Krešimir Friganović, Ph.D. The Chinese researchers working on the projects were Prof. Yueming Gao, Prof. Min Du, Prof. Mang I Vai, Prof. Peng Un Mak, Prof. Sio Hang Pun; Figs. 2 to 7. The group jointly trained 4 Ph.D. students (1 in Croatia, 2 in Macau, 1 in China), and 37 Master students (16 in Croatia, 4 in Macau, and 17 in China).

**Figure 2.** The first meeting: Hangzhou, China at I2MTC 2011; Željka Lučev Vasić (FER) and Yueming Gao (FZU).

Less than two weeks after submitting their first project proposal, Yueming Gao from Fuzhou University and Željka Lučev Vasić from University of Zagreb met for the first time in person, on the outskirts of IEEE International Instrumentation & Measurement Technology Conference (I2MTC 2011) held in Hangzhou, China in May 2011, Fig. 2.

The first work meeting of the newly established collaboration group took place during the IFMBE World Congress on Medical Physics and Biomedical Engineering, held in Beijing, China in May 2012, Fig. 3.

**Figure 3.** WC 2012, May 2012: Sio Hang Pun (UMAC), Peng Un Mak (UMAC), Željka Lučev Vasić (FER), Mario Cifrek (FER), Yueming Gao (FZU), Mang I Vai (UMAC), Pedro Antonio Mou (UMAC).

Further on, the researchers visited each other’s labs on multiple occasions, where they gave lectures and conducted experiments together. In May 2013, an International workshop on intrabody communication was organized in Zagreb, endorsed by IFMBE (International Federation for Medical and Biological Engineering) and IEEE (Institute of Electrical and Electronics Engineers). Members of the research groups from Croatia and China, as well as researchers from FER, presented their work on intrabody communication and signal transmission on/through the human body.

**Figure 4.** Measurements at FER, Zagreb, May 2013; Sio Hang Pun (UMAC), Peng Un Mak (UMAC), Yueming Gao (FZU), Mang I Vai (UMAC), Željka Lučev Vasić (FER)

In June 2015, a delegation of the Chinese Ministry of Science and Technology, led by Deputy Minister Mr. Li Meng, and accompanied by representatives of the Embassy of the People's Republic of China in Croatia led by Ambassador Ms. Deng Ying, visited FER. After the presentation of completed and active bilateral projects between researchers from FER and China, the delegation visited the Department of Electronic Systems and Information Processing and the intrabody communication research facilities, Fig. 5.

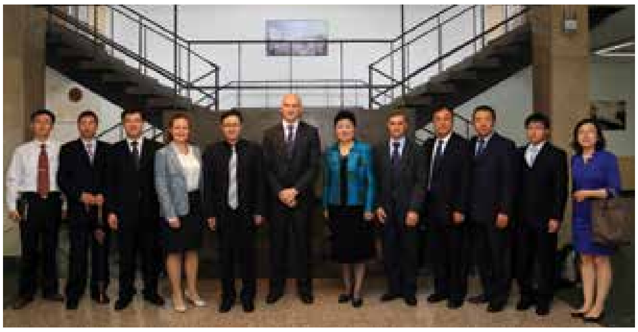

**Figure 5.** June 2015, FER: Yueming Gao (1st), Željka Lučev Vasić (4th), deputy minister Meng Li (5th), vice-dean for science Gordan Gledec (6th), ambassador Ying Deng (7th), Mario Cifrek (8th) with delegation of the Chinese Ministry of Science and Technology and the Embassy of the People's Republic of China in Croatia.

As a consequence of continuous collaboration and common interests in academic, scientific and cultural affairs, in 2018 the International Memorandum of Agreement was signed between the College of Physics and Information Engineering, Fuzhou University and University of Zagreb, Faculty of Electrical Engineering and Computing, with the objective of promoting academic cooperation in education and research.

The collaboration between the Universities of Zagreb, Fuzhou, and Macau was chosen as one of three collaborations between Croatian and Chinese educational institutions to be presented to the prime ministers at the "Exhibition on Cooperation in the Field of Education of Central and Eastern European Countries and China" during the 8th Summit of Central and Eastern European Countries & China in in Dubrovnik, Croatia in April 2019.

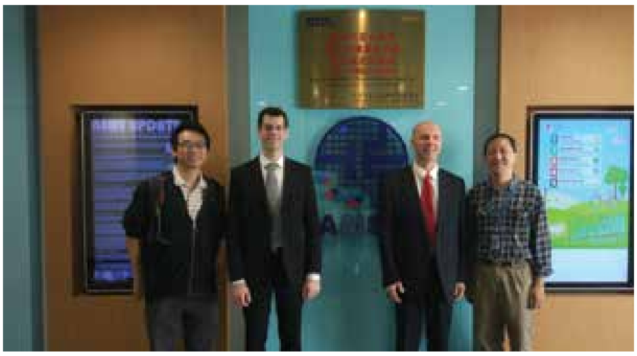

**Figure 6.** December 2017, Macau: Sio Hang Pun (UMAC), Josip Lončar (FER), Silvio Hrabar (FER), Mang I Vai (UMAC).

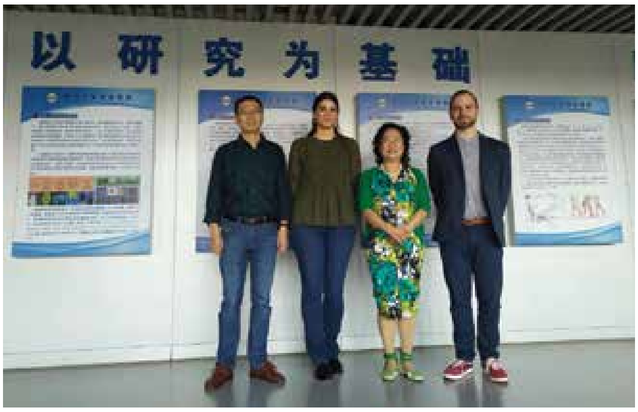


**Figure 7.** October 2019, Fuzhou: Yueming Gao (FZU), Ivana Čuljak (FER), Min Du (FZU), Krešimir Friganović (FER).

At the 2020 IEEE International Instrumentation & Measurement Technology Conference (I2MTC 2020) in May 2020, a special session "Intrabody communication for body area networks" in which 5 papers were presented was organized and co-chaired by Yueming Gao and Željka Lučev Vasić.

As of today, the Croatian and Chinese groups are still working closely together in the biomedical engineering field and several new projects and papers are under the review process.

## Bibliography

Since 2013, when the first common papers were published, in addition to papers published by each group independently, Croatian and Chinese groups have co-authored 11 journal papers. Nine of these papers were published in journals indexed in Web of Science database (1 Q1, 3 Q2, 3 Q3, and 2 Q4 papers). The remaining two papers were published in a new journal not yet included in WoS but already in Q2 quartile in Scimago database. The researchers also participated in 11 conferences with 16 common papers. The common conference paper "An Investigation of the FEM Simulation for the Galvanic Coupling IBC Based on Visible Human Data" received the best paper award at the international conference 2015 IEEE International Conference on Consumer Electronics - China, held in April 2015 in Shenzhen, PR China

**Common journal papers:**

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# 萨格勒布大学电气工程与计算学院与福州大学合作的十年

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[2] 福州大学物理与信息工程学院

2011年2月的一个午后，一位来自福州大学物理与信息工程学院的高跃明博士浏览了中国科技部官方网站的一条项目征集指南。随即他给邮箱通讯录中萨格勒布大学电气工程与计算学院的博士生 Željka Lučev Vasić 发了一份邮件，提议由双方科研导师共同申报克罗地亚-中国科技政府间科技合作项目。当时两个团队正从两条不同的技术路线上开展人体通信技术研究，共同致力于如何将人体本身作为通信信道来实现体表和内部各种生理、健康设备之间的低功耗数据传输。功夫不负有心人。2011年 10 月，合作项目 "体域网中人体通信技术的研究" 获得双方政府资助。该项目负责人是萨格勒布大学 Mario Cifrek 教授和福州大学兼职教授、澳门大学副教授韦孟宇。图1所示，该项目研究人员来自三所高校。

图1 合作单位：福州大学，萨格勒布大学，澳门大学

从那次合作起，克中双方一直保持着良好的合作，包括人体通信以及用于健康和康复监测的无线设备和网络领域。主要的研究课题是电流耦合和电容耦合人体通信系统模型、信道测量以及收发电路设计，最近又扩展了肌阻抗图和运动生理参数监测等领域的研究。

克罗地亚和中国的研究人员具有相同的学术背景和研究方向，但又侧重于相似研究问题的不同方面。克方在电容式人体通信，生物力学中的肌肉疲劳评估、生物医学传感器（如 EMG, ECG, EEG），超带宽(UWB)和生物医学信号处理等领域有丰富的经验。中方(福州大学和澳门大学)擅长于电流耦合人体通信，生物医学系统的电磁建模，多种生化参数的实时监测以及医疗器械的设计。他们也拥有医疗器械开发的产业化和认证经验。迄今为止，合作团队承担了4个双边政府间合作项目1个中国科技部国际合作项目和 2 个福建省科技厅对外合作项目：

1、2011-2013,"体域网中人体通信技术的研究"，双边项目，Mario Cifrek 教授和韦孟宇副教授

2、2013-2015,"体域网中人体通信多耦合技术的合作研究"，中国科技部国际科技合作项目，韦孟宇副教授和 Mario Cifrek 教授

3、2015-2017,"健康物联网中人体通信关键技术的合作研究"，双边项目，Mario Cifrek 教授和高跃明研究员

4、2018-2020,"体内通信技术在健康体域网中的应

用研究”, 双边项目, Željka Lučev Vasić 助理教授和高跃明研究员

5、2018-2021,“基于肌肉阻抗描记法的穿戴式肌肉疲劳程度检测技术合作研究”，福建省科技计划项目，Željka Lučev Vasić 助理教授和高跃明研究员

6、2020-2022,“用于运动疲劳监测的体域网研究”, 双边项目, Mario Cifrek 教授和高跃明研究员

7、2021-2024, “多频激励阻抗谱在腰背肌肉状态动态评估中的合作研究”, 福建省科技计划项目, Željka Lučev Vasić 助理教授和高跃明研究员

克罗地亚在这些项目中的主要研究人员有： Željka Lučev Vasić 助理教授、Mario Cifrek 教授、Igor Krois 教授、Silvio Hrabar 教授、Josip Lončar 助理教授、Ivana Čuljak 博士以及 Krešimir Friganović 博士。中国在这些项目的主要研究人员有：高跃明研究员、杜民教授、韦孟宇副教授、潘少恒副教授、麦炳源助理教授，如图 2 至图 7。双方已经合作培养了 4 名博士（克罗地亚 1 名，中国澳门 2 名，中国内地 1 名）和 37 名硕士（克罗地亚 16 名，中国澳门 4 名，中国 内地 17 名）。

在提交第一个项目提案后两周内，福州大学的高跃明和萨格勒布大学的 Željka Lučev Vasić 于 2011 年 5 月在中国杭州召开的 IEEE 国际仪器仪表与测量技术会议（I²MTC 2011）首次会面，如图 2 。

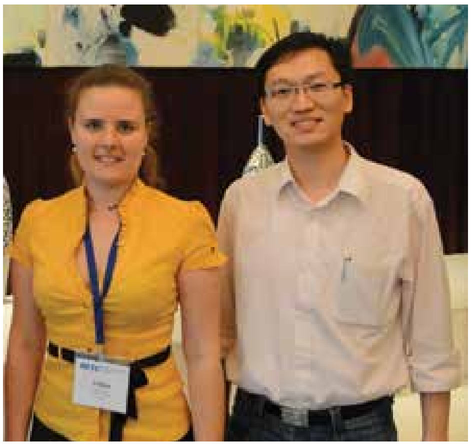

图2 在中国杭州 IEEE国际仪器仪表与测量技术会议的第一次会面，Željka Lučev Vasić（萨格勒布大学）和高跃明（福州大学）

在2012 年5 月中国北京的 IFMBE 世界医学物理与生物医学工程学术大会期间，新成立的项目合作组进行了第一次工作会议，如图 3。

此外，研究人员多次互访实验室，并做演讲和实验。在2013年5月，由 IFMBE 和 IEEE 资助的国际人体通信研讨会在萨格勒布举行。来自克罗地亚和中国的研究组成员，以及萨格勒布大学的研究人员介绍了他们关于人体通信和人体信号传输方面的工作。

图3 2012年5月的会面，潘少恒(澳门大学)，麦炳源(澳门大学)，Željka Lučev Vasić (萨格勒布大学)，Mario Cifrek (萨格勒布大学)，高跃明(福州大学)，韦孟宇(澳门大学)，Pedro Antonio Mou (澳门大学)

图4 2013年5月在萨格勒布大学的实验测量，潘少恒(澳门大学)，麦炳源(澳门大学)，高跃明(福州大学)，韦孟宇(澳门大学)，Željka Lučev Vasić(萨格勒布大学)

2015年6月，科技部李萌副部长在中国驻克罗地亚大使陪同下来访萨格勒布大学。在听完萨格勒布大学与中国研究团队已完成的双边项目的介绍之后，代表团参观了双方在人体通信研究成果并合影留念，如图5。

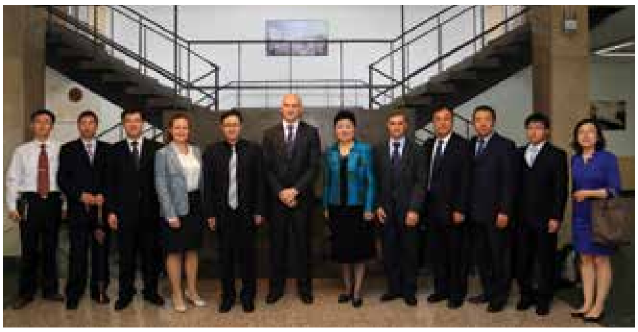

图5 2015年6月中国科技部代表团和中国驻克罗地亚大使馆于萨格勒布大学参观项目组合作情况，高跃明(左一)，Željka Lučev Vasić(左四)，李萌副部长(左五)，Gordan Gledec 副院长(左六)，邓英大使(左七)，Mario Cifrek 教授(右五)

由于在学术、科学和文化事务等方面的共同兴趣和持续合作，秉承着促进教育和研究的学术合作的目

标，福州大学物理与信息工程学院与萨格勒布大学电气工程与计算学院在 2018 年签署了国际合作备忘录。

在 2019 年 4 月中国国家总理李克强先生出席在克罗地亚杜布罗夫尼克举办的第八次中国–中东欧国家领导人会晤，期间参观了“中国–中东欧国家教育合作交流展”。展会上萨格勒布大学，福州大学和澳门大学的合作交流事迹被选为克罗地亚与中国教育机构的三项合作典型之一向克强总理介绍。

在 2020 年 5 月的 2020 年 IEEE 国际仪器仪表与测量技术会议（I2MTC 2020），高跃明与 Željka Lučev Vasić 组织并共同主持了一场“体域网的人体通信”特别会议，并提交了5篇论文。

时至今日，克中双方在生物医学工程等领域仍保持着密切的合作，并且多个新的合作项目和论文正在申报和评审中。

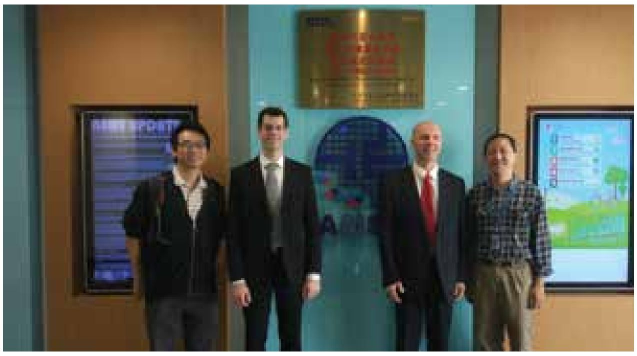

图6 2017 年 12 月在澳门，潘少恒（澳门大学），Josip Lončar（萨格勒布大学），Silvio Hrabar（萨格勒布大学），韦孟宇（澳门大学）

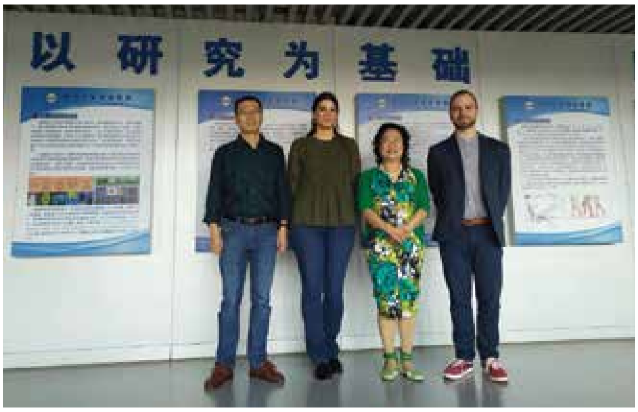


图 7 2019 年 10 月在福州，高跃明（福州大学），Ivana Čuljak（萨格勒布大学），杜民（福州大学），Krešimir Friganović（萨格勒布大学）

参考书目：

自 2013 年首次共同发表论文以来，除各组独立发表的论文外，克中双方已合作发表 11 篇期刊论文，其中 9 篇可在 Web of Science 数据库中检索（1篇 Q1,3 篇 Q2,3 篇 Q3和 2 篇 Q4）。剩下两篇文章发表在一个新期刊上，暂未收录在 WoS 中，但已经包含在 Scimago 数据库的 Q2 中，研究人员还参加了 11 次会议，发表 16 篇会议论文。其中共同发表的论文 “An Investigation of the FEM Simulation for the Galvanic Coupling IBC Based on Visible Human Data” 在 2015 年 4 月在中国深圳举行的 2015 IEEE International Conference on Consumer Electronics - China 国际会议上获得最佳论文奖。

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*Željka Lučev Vasić[1], Yueming Gao[2], Min Du[2], Mario Cifrek[1]*

# INTRABODY COMMUNICATION RESEARCH

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Intrabody communication (IBC) is a new type of wireless communication, in which the human body, together with its immediate environment, becomes a part of a communication channel. IBC systems exploit the electrical properties of the human tissues for the transmission of signals between various wireless electronic devices (transmitters and receivers) placed on the surface of the skin, in its vicinity, or implanted inside the user's body [1-4]. Such devices can be health monitoring devices (heart rate, blood pressure, or body temperature monitors), sensors of physiological signals (electrocardiogram, ECG; electromyogram, EMG; electroencephalogram, EEG), biomedical implants (pacemakers, hearing devices, endoscopic capsules), or devices for assisted living. Typically, communication between wireless devices is accomplished using standard wireless communications, such as Wi-Fi, Bluetooth, Bluetooth Low Energy, RFID, NFC, ZigBee. However, standard wireless communications have not been designed to be used in the vicinity of the human body - they have been optimized for other applications and have either high power consumption, safety issues, or low data rates. As their alternative in the vicinity of the human body, intrabody communication has been proposed. IBC limits the communication range to the user's body, operates at lower frequencies and lower distances than standard wireless systems and accordingly have lower power consumption. Due to the reduced power consumption, heating and tissue irritation of the users are lower, and the battery lifetime is longer. Using IBC also provides an inherent security mechanism: since the communication signal is dominantly confined to the human body, it is difficult to intercept and eavesdrop.

Two main methods of intrabody communication are galvanic and capacitive coupling. In a galvanic coupling method electrodes of IBC devices are in direct contact with the human body. A single signal differential path is established through a current flow that penetrates into internal tissues. In capacitive coupling, a forward signal path is established through the human body and a return path is formed through the environment. This feature allows the interconnection of devices that are both deployed on the same body surface or close to it, without the need for direct contact with the skin. The capacitive method allows higher achievable data rates and lower path loss compared to the galvanic IBC method, especially for higher communication distances on the body. It has been shown recently that a stable capacitive return path can be accomplished even in implantable devices, in case the ground electrode is isolated from the human tissue [4-9]. In [10] the authors analysed compliance of the current density and electric/magnetic fields generated in different modalities of IBC with the established safety standards using the circuit and FEM based simulations. The results show the currents and fields in the capacitive IBC system are orders of magnitude smaller than the specified safety limits. However, galvanic HBC with differential excitation at the wrist can result in localized current densities and field intensities around the electrode, which are significantly higher than the safety limits. They also carried out a small *in vivo* study of vital parameters monitoring using capacitive IBC and the acquired data statistically showed no significant change in any of the vital parameters of the subjects.

The transmission characteristic of an IBC system depends on the properties of tissue and a signal path, which is defined by the signal transmission method, location of the transmitter relative to the receiver, environment configuration, signal amplitude, carrier frequency, and type of modulation. The selection of the appropriate carrier frequency in IBC arises from a trade-off between several factors, like a type of signal coupling, safety regulations to avoid interference with common biological signals, specifications of very low consumption and high tissue conductivity, external noise, and so forth. As opposed to standard wireless systems, which require antennas for communications, IBC systems require only small electrodes. Signal and ground electrodes can be connected to the body, but they can also be left floating, depending on the signal frequency, coupling technique, and application [4].

The latest state of technology related to intrabody communication was published in review papers in 2018 [2] and 2020 [4]. IBC research directions of the research group are IBC channel characterization by means of *in vivo* measurements and modelling, and the development of IBC prototype devices for a specific application.

## Measurements of IBC channel characteristics

In intrabody communication many overlapping physical mechanisms occur at the same time, making channel characterization and measurements a challenging task. In addition to this, IBC channels change dynamically with electrode positions and size, subject, subject's movements, and surrounding environment.

Establishing a proper procedure and measurement setup for measuring IBC channel characteristics, while keeping the overall IBC signal path intact, is a very challenging task [11, 12], since introducing any kind of measuring equipment into the IBC channel modifies the return signal path and influences the measurement results. For accurate measurements, measuring equipment (signal generator, oscilloscope, network and spectrum analyser) should be galvanically decoupled from the IBC channel. This is usually achieved using an optical link, differential probe or, more often, connecting balun transformers between the transmitter/receiver electrodes and the rest of the measuring equipment, as in Fig. 1 [4, 13, 14].

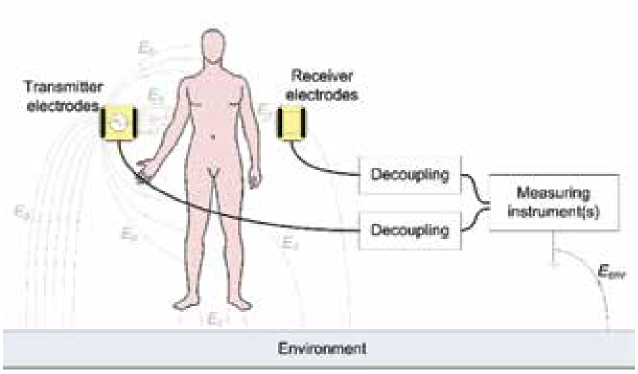


**Figure 1.** General capacitive IBC measurement setup using commercial equipment.

The group investigated influences of the type, size, and position of the transmitter and receiver electrodes, and the influence of the environment in a capacitive IBC channel. Several thousand *in vivo* measurements were performed on a larger number of test subjects with different anatomical characteristics, for several static and dynamic body positions in the frequency range from 100 kHz to 100 MHz. Channel gain was measured using commercial network analysers (power-line and battery powered), with and without balun transformers for decoupling. Transmission characteristics of the capacitive intrabody communication channel in all measuring combinations showed band-pass characteristics: an increase 20 dB/decade up to around 45 MHz, and a steep decrease at higher frequencies, as in Fig. 2.

Additional measurements were performed using a proprietary battery-powered transmitter for signal generation, and the battery-powered spectrum analyser for measuring the received signal power [15]. The results agree qualitatively with the previous ones.

However, it has been recently shown that the value of the capacitance between primary and secondary windings of the transformer and its symmetry with

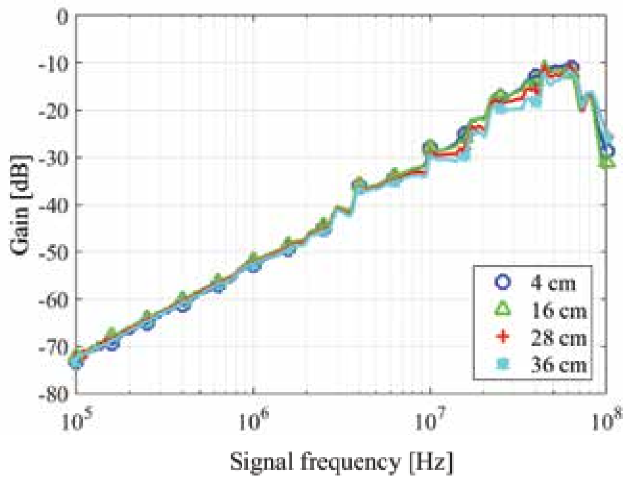


**Figure 2.** Capacitive IBC channel transmission characteristics measured at four transmitter-receiver distances.

respect to the ground could influence results drastically [14, 16]. Furthermore, using balun transformers and commercial equipment with 50 Ω input impedance results in higher measured gain than in a realistic IBC channel due to the improper ground isolation, and with lower gain at low frequencies due to the low frequency termination [17, 18]. Also, devices with large physical size (like commercial network and spectrum analysers) create a larger than expected return path, whether they were isolated with baluns or not, thereby increasing the measured channel gain [4, 16-18]. Therefore, for performing accurate measurements of IBC channel transmission characteristics, testing apparatuses should be of the same physical size and have the same grounding configurations as devices that will eventually be employed in IBC applications, with the corresponding matching networks between devices and the human body. In other words, measurements of any IBC channel transmission characteristics should be performed using small and independent battery-powered devices, thus bypassing the need for galvanic decoupling and providing a more realistic IBC channel.

Currently, the group is developing proprietary small battery-powered devices (signal generator and received power meter) for IBC channel characterization in a realistic communication scenario and a wider frequency range. The plan is to use them for IBC channel gain measurements for devices worn on the human body, and also for the implants.

Preliminary results of the first *in vivo* measurements of capacitive intrabody communication with implant-like devices on humans were presented in [19]. The IB2OB channel was mimicked by placing the transmitter under the armpit and taking different body positions while covering transmitter electrodes with tissue. The results agreed qualitatively to the results of the on-body channel measurements obtained using the same battery-powered equipment and baluns for decoupling as in [13, 14].

Since experiments with implanted capacitive IBC devices on living beings would be highly invasive, measurements of implantable capacitive IBC channels are usually made on human body phantoms rather than on humans. Human tissue phantoms are made by combining simple chemical substances with water for adjusting the conductivity and relative permittivity of the solution. However, it is rather difficult to produce solutions that emulate the electrical properties of human tissues in a wide range of frequencies, so a single phantom can be used at a specified frequency or in a narrow frequency band. The tissue phantom in which the in-body transmitter is placed needs to be liquid, so the distance between transmitter and receiver electrodes can be adjusted during the measurements; while outer tissues in multilayer phantoms can be semi-solid or animal skin. Receiver electrodes are placed inside the phantom for implantable to implantable channel measurements, and on the outer layer of a phantom for implantable to on-body (IB2OB) channel measurements, as in Fig. 3. Transmitter electrodes are immersed in the liquid phantom and their position can be adjusted in in all three directions. The liquid phantom in Fig. 3 has conductivity similar to human muscle tissue (7.38 mS/cm @ 22.7 °C), which is achieved adding 56 g of sodium chloride to 14 l of distilled water. The first results obtained using the aforementioned setup are promising.

**Figure 3.** Measurement setup for implantable to on-body channel measurements on a muscle tissue equivalent liquid phantom.

## IBC channel characterization by modelling

The research group developed several types of IBC channel models of the human limbs, most recent based on anthropometric data of several persons [20] and based on Visible Human Data [21].

In [20] the safety of galvanic IBC was analysed using empirical FEM arm models based on the geometrical information of six subjects with different physiological characteristics. The weight, fat percentage, and muscle percentage for each subject were measured and geometrical dimensions for the arm model in Fig. 4 were calculated and models were developed in COMSOL 5.2 Multiphysics Software. The electric field intensity and localized SAR were computed and, in some cases, 2010 ICNIRP safety limits were exceeded. To comply with safety standards, the use of a frequency signal of 100 – 300 kHz has been proposed for galvanic IBCs, allowing a current signal of 1–10 mA and a voltage signal of 1–2 V.

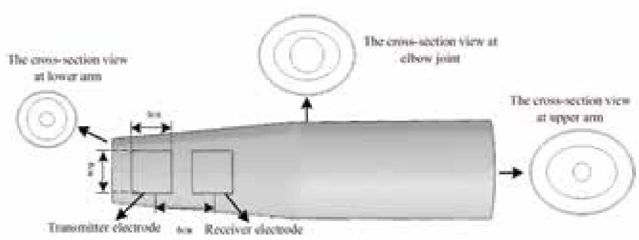


**Figure 4.** The empirical equivalent arm model and the electrode configurations, [20].

Visible Human Data (VHD) set [21] contains transverse anatomical images of a male taken in cross-sections 1 mm apart and showing all internal tissues. VHD images include different textures, densities, colours, and other details that are difficult to reconstruct in the 3D layer model directly, so the modelling was implemented using several software packages, such as Photoshop, Mimics, Geomagic Studio, Solidworks. The original images were firstly divided into tissue layers (skin, fat, muscle, and bone, if necessary). The outlines of each tissue were extracted automatically on every anatomical image and the contour lines of tissues were reconstructed using 3D reconstruction software, to better differentiate each layer. Contours were filled with the respective tissue and a 3D model was smoothed and divided into 3D models of each tissue. Tissue conductivity $\sigma$ and permittivity $\varepsilon$, were derived from the Gabriel parametric models [22].

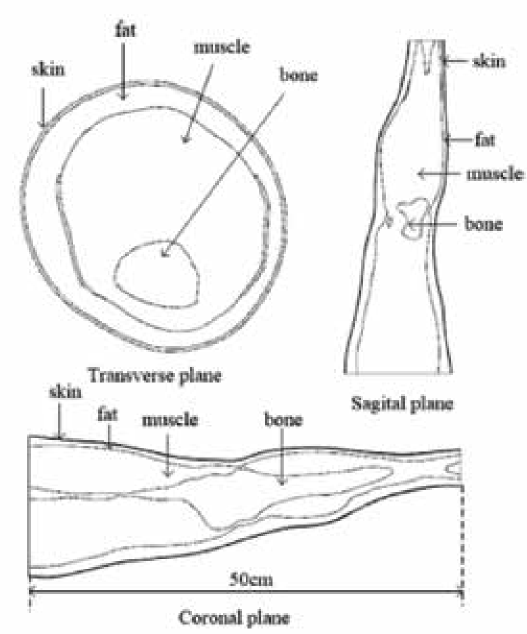


**Figure 5.** Human body leg model based on Visual Human Data, [23].

The numerical leg model based on Visual Human Data consisting of skin, fat, muscle, and bone layers is presented in Fig. 5, [23]. The model was used for simulation of galvanic IBC communication between implanted medical devices and external equipment in a frequency range between 10 kHz and 1 MHz, [23]. Transmitter electrodes were attached to the skin surface and the receiver electrodes were placed between the muscle and fat layers. The transmitter-receiver distance was set to 6 cm or 30 cm near the ankle, knee, and hip, respectively, e.g. in Fig. 6.

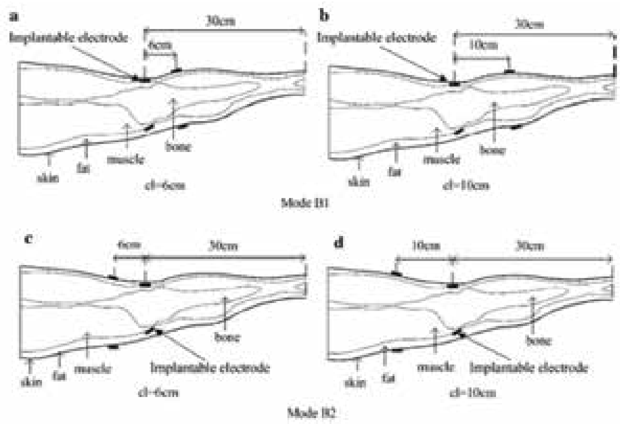


**Figure 6.** Several positions of electrodes near the knee, [23].

Additionally, two layer phantom models of a leg were built with tissue conductivities and permittivities chosen for 40 kHz frequency. Since at 40 kHz the conductivities of the fat and skin layers were almost the same and the influence of the bone tissue is minor, the model consisted of a muscle and a skin-fat layer. Both tissues were made by mixing agar, potassium chloride, hydroxyethyl cellulose (HEC) and distilled water. The electrical conductivity of the layer was set adjusting the quantity of potassium chloride in the mixture. Outer contours of the muscle and skin-fat layers based on the Visible Human Leg data were 3D printed and used as a mould for the mixture. After the layers of the model were produced, electrodes were placed at the same positions as in the numerical model, Fig. 7. A detailed explanation of the design, production, and verification of the phantom model can be found in the paper [24].

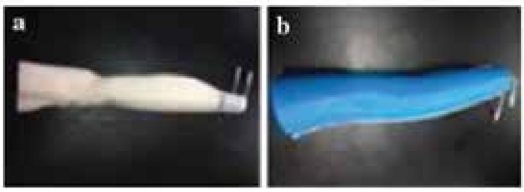


**Figure 7.** Phantom model, (a) muscle layer (white), (b) complete model, outer is skin-fat layer (blue) [23].

The transmission characteristics were calculated and measured for numerical and phantom models, respectively. Both models showed the same characteristics for all electrodes positions, which proved that the simple phantom model can be used as an effective supplement to the FEM model in the design and performance test of implantable transceiver, as well as in the research of implantable channels in the future.

## IBC devices

In the design of IBC devices, the design of the matching network and the choice of the optimal modulation method are important aspects.

Programmable System-on-Chip (PSoC) is a family of microcontrollers which include a CPU core and mixed-signal arrays of configurable integrated analog and digital peripherals that can be arbitrarily routed and interconnected. IBC systems based on PSoC microcontrollers were developed for low data-rate applications. Transmitters acted as signal generators and synthesized a continuous FSK (frequency shift keying) [25], BPSK (binary phase shift keying) [26], or on-off-keying (OOK) [27] modulated signals using digital to analog conversion in the microcontroller. The receivers performed the demodulation and recovering of the sent digital data. The developed systems were tested *in vivo* and successfully achieved the desired functionality, especially considering that no external components were added in the systems [25] and [26] other than the electrodes, and only two passive external components were added to the system [27], Fig. 8. Methods for increasing generated signal frequency up to several megahertz on PSoC platform will be explored and other modulation methods will be tested in order to find optimal communication requirements for PSoC platform.

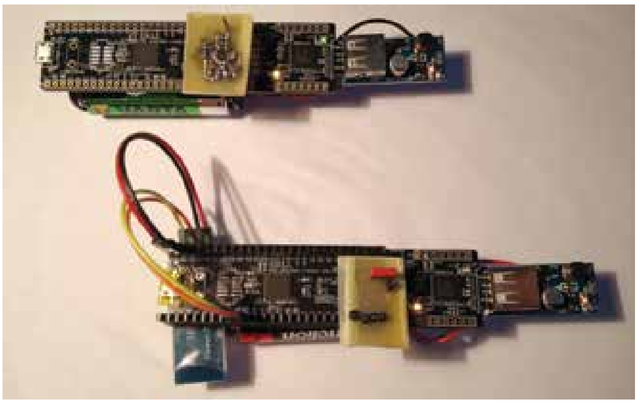

**Figure 8.** OOK receiver (up) and transmitter (down) on PSoC boards described in [27].

The development and improvements of the IBC interface and FPGA transceivers for galvanic IBC devices were described in papers [28-30]. In [28] field programmable gate array (FPGA, XC6SLX16) was used as a platform for testing modulation and demodulation methods in different application scenarios. Direct sequence spread spectrum (DSSS) communication

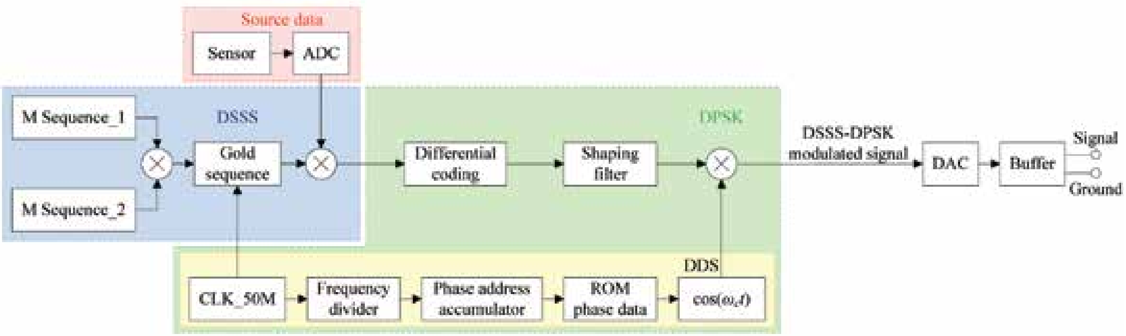


**Figure 9.** Block diagram of DSSS-DPSK transmitter, [27].

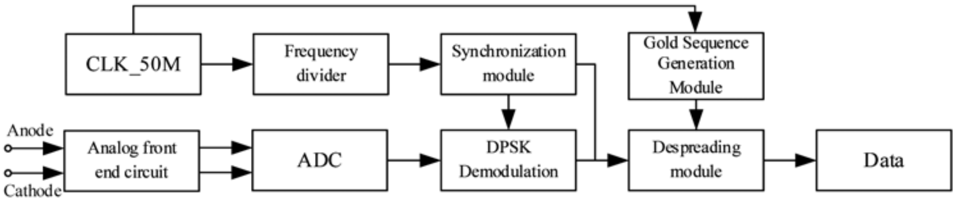


**Figure 10.** Block diagram of the receiver, [27].

and phase modulation were adopted to realize DSSS-differential phase shift keying (DPSK) and DPSK modulation transmission of baseband data. The block diagram of DSSS-DPSK transmitter is presented in Fig. 9. The transmitter is composed of a source module, a direct digital synthesis (DDS) module, a spread spectrum module, and a DPSK modulation module. The DSSS-DPSK signal is sent to the DAC (digital-to-analog converter) and then to the buffer. Finally, the signal is sent to the human body for transmission via transmitter signal and ground electrodes. The DPSK transmitter was also built equipped with the same peripheral circuit. The overall design of the DSSS-DPSK signal receiver is shown in Fig. 10. The main parts of the receiver are an analog front end (AFE), a DPSK demodulation module, a despreading module, and a synchronization module. The analog front end mainly preprocesses the signal that enters the receiver. The Costas loop method was employed to achieve reliable symbol recovery. *In vivo* experiments were conducted to compare the performance of DSSS-DPSK and DPSK galvanic coupling IBC transceiver systems under the same conditions. The generated signal frequency was 2 MHz for both DSSS-DPSK and DPSK transmitters. The set channel lengths (transmitter-receiver distances) were 10 cm, 30 cm, 90 cm, and 120 cm. The influence of human activity (arm still or moving), signal-to-noise ratio (SNR), and transmission distance were tested and compared by measuring the bit error ratio. The bit error ratio (BER) was calculated dividing the number of bits received in error by the total number of bits transmitted within the same time period:

$$BER = \frac{\text{Number of bit errors}}{\text{Total number of bits}},$$

so the lower the BER value the better.

In order to test the BER performance, two PCIe-6361 data acquisition cards were used to collect the baseband data at the transmitter and the demodulated data at the

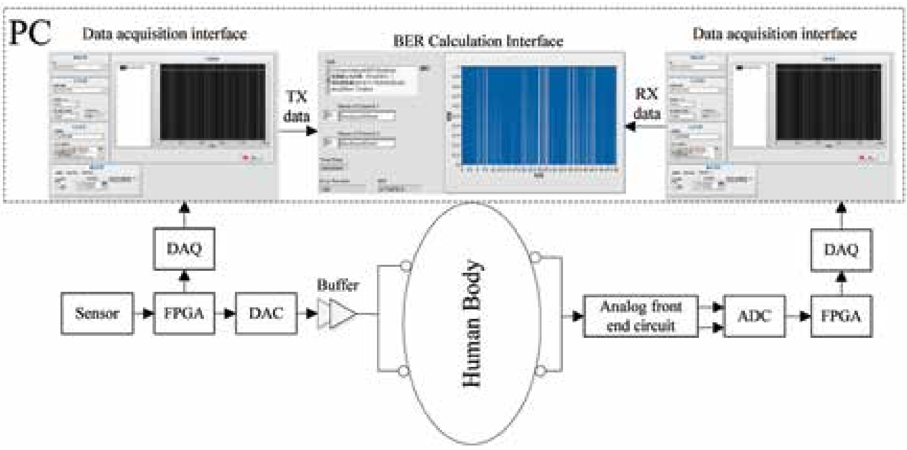


**Figure 11.** BER test platform

receiver, as in Fig. 11. As expected, the BER decreased as SNR improved for both cases, regardless of the body movements of the test subject, Fig. 8. BER was also lower at lower transmitter-receiver distances. It was shown that DSSS-DPSK modulation requires a lower SNR than DPSK modulation. The BER measured with DPSK transceivers was 40 times greater than with DSSS-DPSK transceivers at a transmitter-receiver distance of 30 cm and different SNR values. When changing the BER from extremely poor ($1.40\times10^{-1}$) to excellent ($1.51\times10^{-6}$), the SNR of DSSS-DPSK transceivers only had to be improved by 16 dB. In contrast, when the BER was changed from extremely poor ($1.54\times10^{-1}$) to good ($1.65\times10^{-5}$), the SNR of the DPSK method had to be improved by 25 dB. With a SNR equal to 5 dB, the BER ratio using DPSK transceivers was 7 times larger than using the DSSS-DPSK transceivers. However, DSSS-DPSK transceivers were statistically more sensitive to changes in motion status than DPSK.

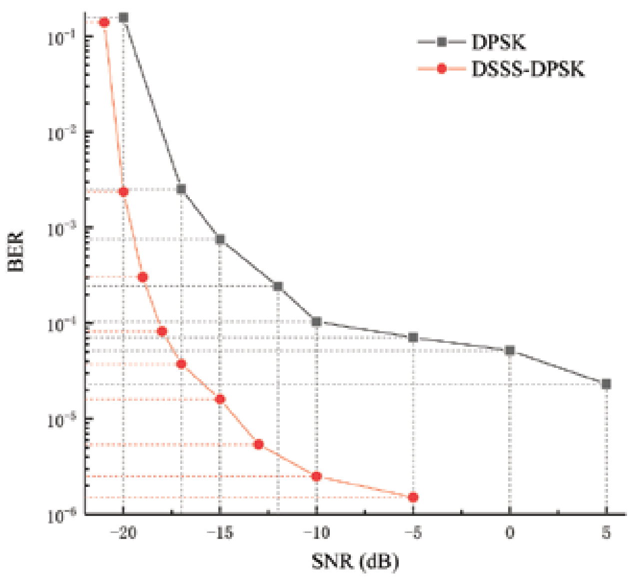


**Figure 12.** Average BER of motion and stationary vs. SNR for both tested modulations.

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# ELECTRICAL IMPEDANCE MYOGRAPHY FOR MUSCLE FATIGUE MONITORING

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Since 2018 the group has been working on the project “Evaluation of the Local Muscle Fatigue with the EIM Method for Wearable Applications” with the aim of developing a small wearable device for muscle fatigue monitoring.

Muscle fatigue is a phenomenon which often occurs in daily life, due to repetitive muscle flexions or long-term static contractions. It occurs when the muscle motor system cannot maintain the expected intensity required for a particular activity due to the weakening of its functional ability and is generally reversible. The continuous movement of muscles gradually reduces their work capacity, maximum contraction force, and output power [1-3]. If muscle fatigue is not properly handled, it can cause muscle strain and seriously affect the daily lives of people or, especially, the physical exercise of athletes. Muscle fatigue can be evaluated by several standard indicators, such as muscle oxygen saturation, lactic acid concentration, ultrasound image entropy, and surface electromyography (sEMG), which is most widely used [1]. However, most of these techniques are conducted in hospitals or rehabilitation centres and the patients cannot detect muscle fatigue anytime and anywhere or perform self-measurement at home. A wearable device that can monitor muscle fatigue at any time would be very useful for exercise rehabilitation, muscle disease diagnosis, sports training, and other fields.

Electrical impedance myography (EIM) is a non-invasive bioelectrical impedance technique based on the four-electrode array [4]. This technique evaluates the health status of a local muscle by applying a high-frequency, low-intensity alternating current to the muscle of interest through the outer electrodes and measuring voltage between the inner electrodes of the array, Fig. 1. EIM can also be used in clinical diagnosis and efficacy evaluation of various neuromuscular diseases. During muscle fatigue, the lactic acid content of muscle cells in muscle fibres increases, thus slowing down the conduction speed of electrical signals in muscle fibres. The EIM method detects changes in impedance due to muscle abnormalities or muscle fatigue, so it can be used for the assessment of muscle fatigue. Compared to the traditional sEMG approach for muscle fatigue assessment, the detection parameters (resistance, reactance, phase) of EIM signals have many advantages: large EIM signal amplitude, controllable frequency, and simple pretreatment procedure. Therefore, EIM could be used as a new, low complexity and high feasibility method of real-time muscle fatigue monitoring, which can easily be integrated with various wearable devices.

Papers [1], [2] and [3] are related to the use of electrical impedance myography for the assessment of muscle fatigue on small wearable devices. In [1] the researchers described and developed a new method for estimating muscle fatigue from signals measured using four-electrode electrical impedance myography. Positions of four electrodes were optimized in [2] and a new arrangement of electrodes was proposed, optimal for estimating muscle fatigue. The new method was used to estimate muscle fatigue during static and dynamic contractions in [3]. The method for estimating muscle fatigue was modelled theoretically and by simulations and verified by *in vivo* measurements. Finally, in [5] electrical impedance myography was used for evaluating muscle fatigue induced by neuromuscular electrical stimulation.

In [1], based on the anthropometric parameters of eight volunteers, dimensions of an equivalent three-dimensional (3D) model of the entire arm were calculated. A standard four-electrode array was placed on the upper arm, above the *biceps brachii* muscle, as in Fig. 1. The model was developed in AC/DC module

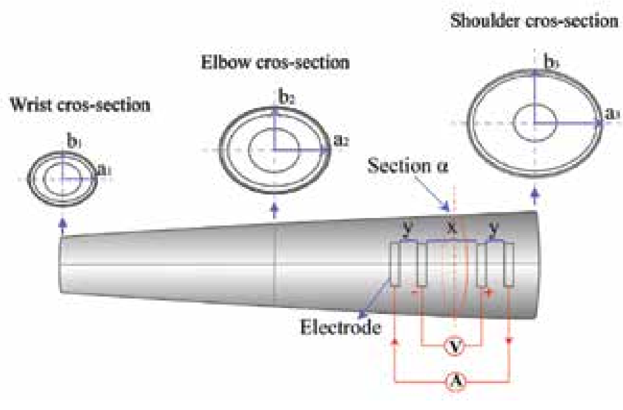



**Figure 1.** Human body 3D arm model has four layers (bone, muscle, fat, and skin), section α represents the cross-section at the midpoint between the excitation electrodes, [1].

of COMSOL Multiphysics 5.2a, a simulation software based on the finite element method (FEM), and used to find the optimal distance between EIM electrodes. An optimal electrode configuration allows a minimal excitation signal for a maximal measured potential difference and provides a good precondition for the front-end signal detection system of a wearable device.

A 1 mA 50 kHz current signal was used as excitation. The overall current density generated at the excitation (outer) electrodes and applied to the arm $J_{arm}$, current density in muscle tissue $J_{muscle}$, and voltage $V_{sense}$ between the inner electrodes were calculated. From the perspective of a wearable device, a higher potential difference between the excitation electrodes leads to a higher voltage detection at the receiving end. In addition, if the signals injected into the human body can flow through the muscle layer, then the measured EIM parameters can accurately reflect the muscle characteristics. Therefore, the distance between EIM electrodes was optimized based on two parameters, which should be as high as possible: 1) the ratio of current density in the muscle layer and the current density in the whole arm ($J_{muscle}/J_{arm}$) and 2) the potential difference achieved at the voltage electrodes ($V_{sense}$). In EIM the distance between the electrodes is mainly set in three ratios of $y$:$x$:$y$ in Fig. 1: 1:1:1, 1:2:1, and 1:3:1, depending on the size of the *biceps brachii* muscles of healthy adults. In this research, the distance between the two internal EIM electrodes was set to standard $x = 24$ mm, as shown in Fig. 1. The distance between the outer and internal electrodes ($y$ in Fig. 1) was set to 8 mm, 12 mm, and 20 mm. Although according to the ratio of 1:1:1, $y$ should be set to 24 mm, the maximum value of $y$ was set to 20 mm because of the limited length of a volunteer's *biceps brachii* muscles.

It was shown that at a frequency of 50 kHz, the current flowing through the muscle layer accounts for more than 90% of the overall current in the arm. The value of $J_{muscle}/J_{arm}$ increases with the space between outer electrodes, while the modulus of voltage $V_{sense}$ decreases. The difference between the shortest (8 mm) and longest (20 mm) distances in terms of $J_{muscle}/J_{arm}$ is less than 1 %, and the modulus of $V_{sense}$ parameter is the largest for $y$ = 8 mm. Therefore distances $y$ = 8 mm and $x$ = 24 mm were found optimal and used for *in vivo* EIM measurements during dynamic muscle contractions.

Before the *in vivo* measurements, the highest load a human arm can hold (maximal voluntary contraction, MVC) of eight volunteers was measured using multi-joint muscle strength assessment training system Biodex System 4, Table 1. In the *in vivo* experiments, different contraction strengths (such as 20% MVC, 40% MVC, and 60% MVC) were used to indicate different contraction states of muscles.

| Volunteer | 1 | 2 | 3 | 4 | 5 | 6 | 7 | 8 |
|---|---|---|---|---|---|---|---|---|
| MVC, [Nm] | 28.7 | 30.2 | 36.4 | 29.1 | 57.0 | 48.8 | 64.5 | 62.0 |

A block diagram and photography of a lower arm flexion experiment are shown in Figs. 2 and 3, respectively. The EIM measurement setup consisted of a signal generator (Rigol DG4162), constant current source, four electrodes placed on *biceps brachii* muscle, Agilent 1141A differential probe, and Agilent MSO7054A oscilloscope. In the experiment, the AC voltage signal with a frequency range from 1 kHz to 1 MHz and 1 V amplitude was generated and fed to the custom-built 1 mA constant current source. The current signal was then loaded into the *biceps brachii* muscle through four side-by-side electrodes of the same size (40 mm x 10 mm). The coupling voltage signal measured between the voltage electrodes at the receiving end was displayed by the oscilloscope in real-time to obtain the impedance parameters. The Agilent 1141A differential probe was used for connecting the electrodes on the body and the oscilloscope in order to solve the common ground problem between the receiving and the transmitting electrodes.

Simultaneously, real-time surface EMG signals were acquired using the Trigno Lab wireless surface EMG acquisition system. The measured sEMG signals were used to verify  of EIM muscle fatigue estimation, since median frequency (MF) of power spectra of sEMG signals is often used as the parameter for muscle fatigue evaluation.

Volunteers were asked to perform repeated lower arm flexions until exhaustion while holding a dumbbell with weights of 20%, 40%, and 60% of their MVC. One dynamic contraction cycle started with the arm naturally drooping at an elbow angle of 180°. During one contraction cycle the elbow angle decreases from 180° to 45° and then increases back to 180°, as in Fig. 2. The EIM parameters were sampled every 10 cycles during this muscle fatigue process at the half-cycle point (45°).

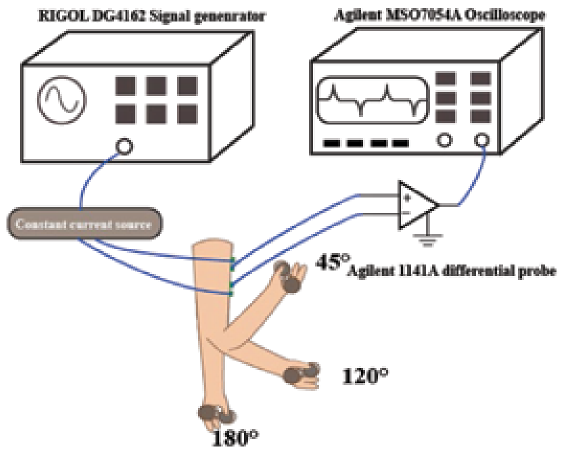



**Figure 2.** Block diagram of a lower arm flexion experiment, [1].

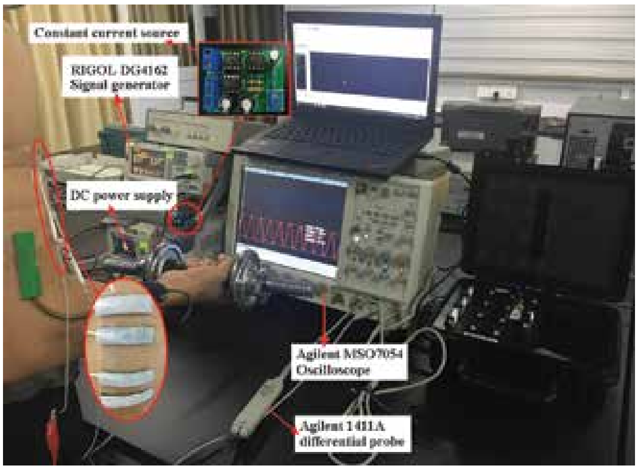



**Figure 3.** EIM human experiment measurement setup, [1].

In many EIM studies, resistance *R* is the first parameter to change during muscle contraction [6, 7] while reluctance and phase experience minimal changes. Therefore, only resistance *R* was used for muscle fatigue evaluation. The results were compared to median frequency of sEMG signals, which is standardly used for muscle fatigue evaluation.

Fig. 4 shows the relationship between EIM *R* parameters and contraction time of *biceps brachii* muscle under different loads (20%, 40%, and 60% MVC) at 50 kHz measured on the volunteer 1. The muscle *R* data at different load levels show a linear decreasing trend with increasing contraction time. Moreover, it is evident that the slope of the linear fit line is different for different loads. The average and standard deviation of measured resistance *R* on eight volunteers were calculated under different load levels and the results are shown in Fig. 5. The values of parameter *R* before (without fatigue) and after muscle fatigue experiment (exhaustion) were significantly different ($p < 0.01$). The difference in *R* before and after the experiment (drop-out value) was about 8 Ω.

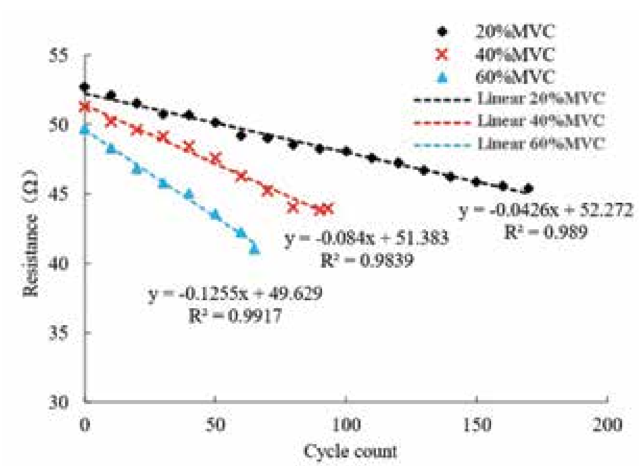



**Figure 4.** Variation of EIM resistance parameters with time at 50 kHz for three different contraction forces, [1].

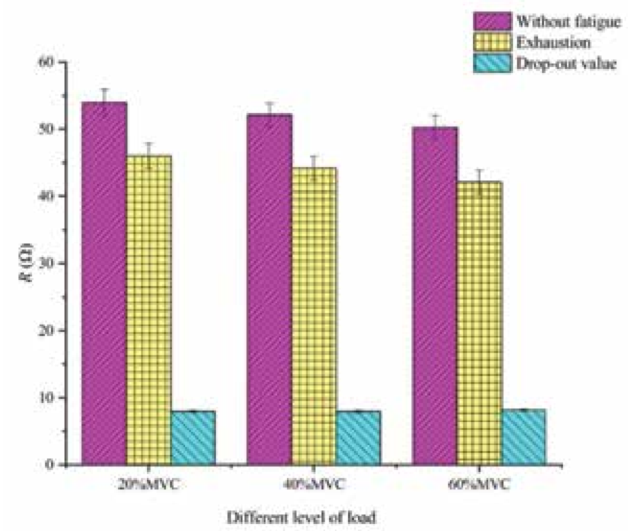



**Figure 5.** Average value and standard deviation of EIM resistance measured on eight volunteers during fatiguing dynamic contractions, [1].

The measured sEMG median frequency (MF) for different loads also follows the same downward trend as EIM *R* parameter, i.e. its value decreases as fatigue level increases, and a higher load results in a more rapid decline.

EIM and sEMG signals were acquired simultaneously during dynamic contractions until exhaustion. The achieved number of contractions cycles decreased with increasing load. Muscle fatigue index (MF) calculated from sEMG signals (blue) and *R* measured using EIM electrodes (red) for 40% MVC load normalized to the number of cycles are shown in Fig. 6. It has been experimentally shown that the absolute values of the linear fit slope are higher for the heavier load and the decline slope of *R* linear fit line for each load is nearly 2/5 that of the MF, [1]. Thus, the decreasing rate of the *R* is not as fast as a decrease in MF, but a considerable

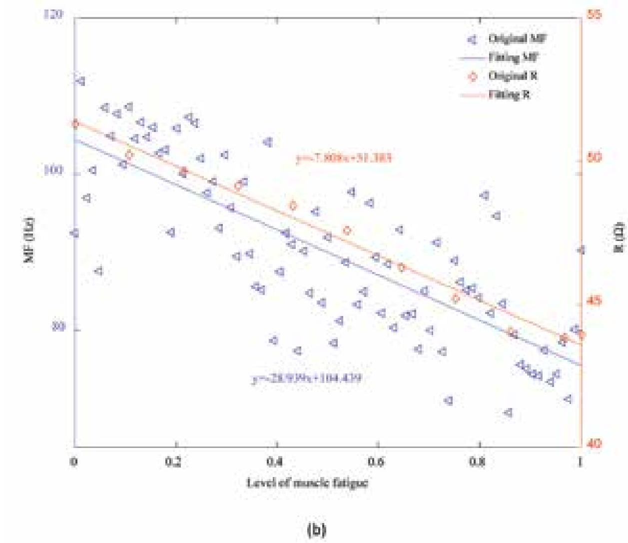



**Figure 6.** Median frequency (MF, blue) and impedance *R* (red) measured during dynamic contractions for 40% MVC, [1].

consistency is observed in the decline regularity of the two curves. Therefore, EIM with a standard electrode four-in-line arrangement can be used for muscle fatigue evaluation, and the rationality of this approach was verified. In the muscle dynamic contraction fatigue experiment, the subjects were considered to reach the semi-fatigue point when the measured *R* parameters decreased by approximately 4 Ω. With the further *R* decrease approaching 8 Ω, the muscle fatigue is considered to reach its limit. Thus, following the magnitude of *R* decline the subjects can adjust their exercise intensity and avoid muscle fatigue or the damage caused by excessive exercise.

After proving the feasibility of exploiting EIM for muscle fatigue evaluation, the researchers further investigated and optimized influential parameters of EIM (electrodes arrangement and distance) based on equivalent circuit [2] and finite element [3] models.

The standard four-in-line EIM electrode arrangement has low potential sensitivity and occupies a large area, which is not suitable for the miniaturization and portable design of wearable devices, so in [2] a new more compact electrode arrangement was proposed, Fig. 7. The surface of electrodes in both models is the same, but their shape is different. In model A, standard 40 mm x 10 mm rectangular electrodes are used arranged in a serial array. In model B 20 mm x 20 mm square electrodes are used, arranged in a parallel array. An equivalent circuit model of both models was proposed and the influence of distances between the electrodes of model B was investigated. The results were tested *in vivo* on six volunteers using Imp™ SFB7, a device for measuring the response voltage while generating a 2 mA current signal.

Two sets of static experiments were performed on six volunteers. In the first set, measuring electrodes were placed on a biceps muscle separated by $Y$ = 24 mm, and excitation electrodes were moved from the position $X$ = 5 mm to $X$ = 20 mm with a 5 mm step. The maximum impedance was measured for $X$ = 5 mm, so in the seconds set of the static experiments, distance $X$ was fixed to 5 mm, and distance $Y$ was changed between 5 and 45 mm with a 5 mm step. In this experiment the muscle electrical impedance increased as $Y$ increased when $X$ was fixed, for all volunteers. Therefore, from the perspective of miniaturization and wearability, the optimal configuration of model B presented in Fig. 7 right is using 20 mm x 20 mm electrodes and spacings $X$ = 5 mm and $Y$ = 24 mm, in parallel array.

The second set of *in vivo* experiments involved measurements of muscle impedance measured using optimal model A and model B configurations in Fig. 7 during dynamic contractions (lower arm flexion, as in Fig. 2). Dynamic biceps contractions were repeated until exhaustion, when the subject cannot complete the movement. For each model, impedance was measured during the movements and the change of the impedance $\Delta R$ between the rest muscle state (beginning of the first cycle) and exhausted muscle state (end of contractions) was calculated. For both models impedance consistently decreased with time, i.e. with increased muscle fatigue. Between the initial state and the last contraction, the decrease of the measured impedance $\Delta R$ was 4–7 Ω for the optimal configuration of model A and the decrease measured by the optimal configuration of model B was 8–14 Ω, depending on the volunteer, as presented in Fig. 8. The experimental results show that the $\Delta R$ measured by the optimal configuration of model B is twice that measured by the optimal configuration of model A, so model B configuration can be used for EIM-based muscle fatigue detection

Further investigations on the optimal model B (parallel) configuration were performed in [3], by means of

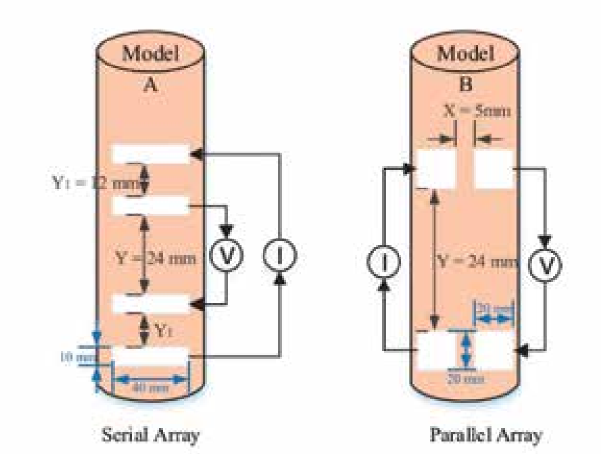



**Figure 7.** EIM electrode configurations based on the four electrodes. Model A is the traditional four-electrode configuration with a serial array. Model B is a proposed four-electrode configuration with a parallel array, [2].

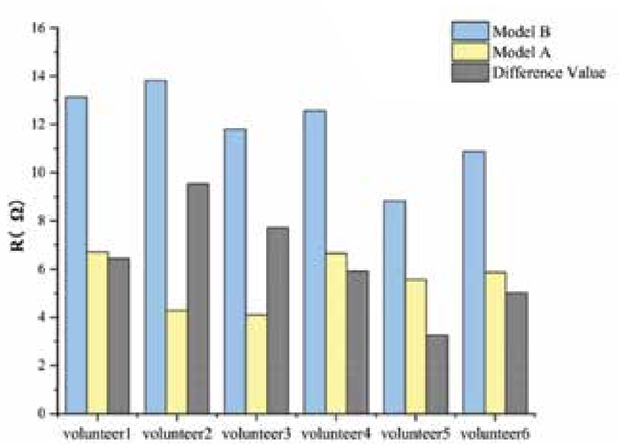
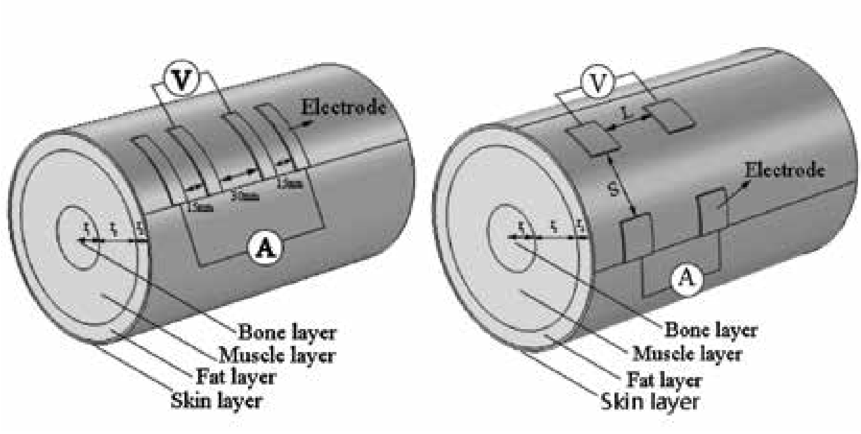



**Figure 8.** Decrease of muscle electrical impedance for models A and B. Difference Value is the difference of variation in muscle electrical impedance between models A and B when muscle is tired, [2].



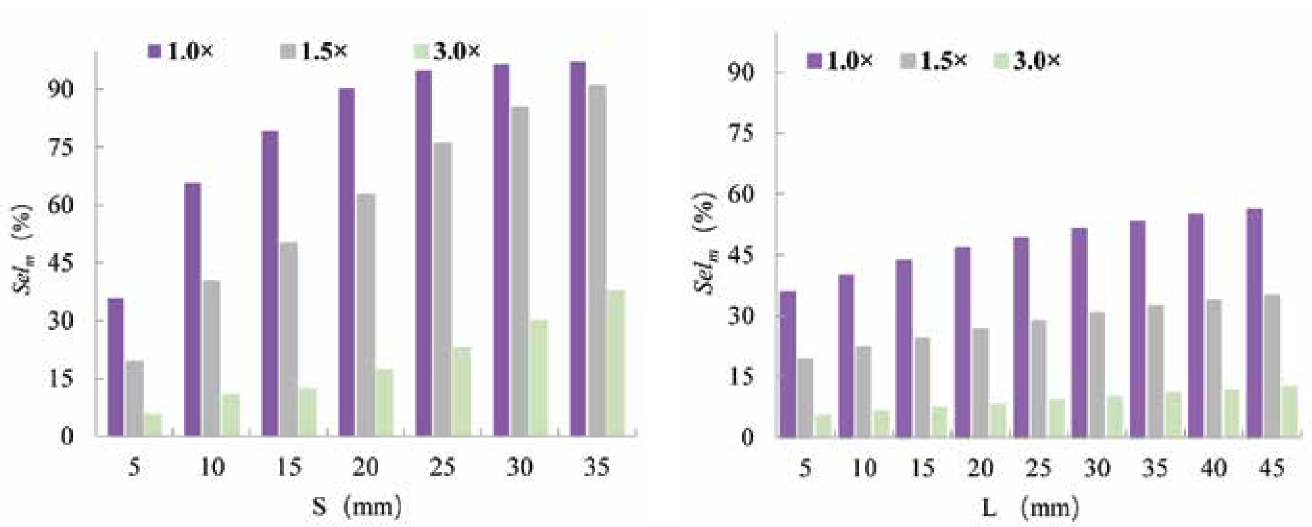

(a) (b)

**Figure 9.** A four-layer FEM model of human upper arm: (a) serial electrode configuration method; (b) parallel electrode configuration method, [3].

numerical simulations in COMSOL and *in vivo* measurements. A four-layer FEM model consisted of skin, fat, muscle, and bone layers, Fig. 9, and was based on anthropometric characteristics of 10 volunteers. Comparing serial and parallel configurations with the equal electrode area and including approximately the same muscle area, it was again proven that parallel configuration results in a larger EIM impedance amplitude at all tested frequencies (10, 25, 50, and 100 kHz). The larger impedance implies a higher detection amplitude, which is easier to integrate into wearable devices.

Apparent impedance is the electrical impedance measured on the skin surface. Its value depends on the contribution of each layer below the electrodes, which depends on the electrode characteristics (size, material, shape, and position on the skin). Sensitivity analysis was used to calculate the impedance of each layer and optimize positions of the four electrodes on the skin surface (distances $S$ and $L$ in Fig. 9) in order to maximize the percentage of muscle tissue contribution to the apparent impedance. It was shown that by adjusting the electrode spacing on the skin surface, the contribution of the muscle tissue in the apparent impedance changes. At 50 kHz, when $S$ = 5 mm and $L$ = 5 mm, the contribution of the fat layer in apparent impedance was 81.05%, and the muscle layer was 19.18%. However, for larger electrode spacing $S$ = 20 mm and $L$ = 20 mm, the contribution of the fat layer decreased to 28.67%, and the contribution of the muscle layer increased to 71.68%. For the bone and skin layers, the difference was small and changed only within 1%.

In order to reduce the influence of individual fat differences on EIM results, the rate of change of EIM impedance with fat thickness was studied for different electrode distances. Fat thickness was set to 6 mm (base value, 1.0x) and thickened with a step size of 3 mm. The results for the fat thickness of 6 mm (1.0x), 9 mm (1.5x) and 18 mm (3.0x) are shown in Fig. 10. The selectivity of impedance to muscle tissue $Sel_m$

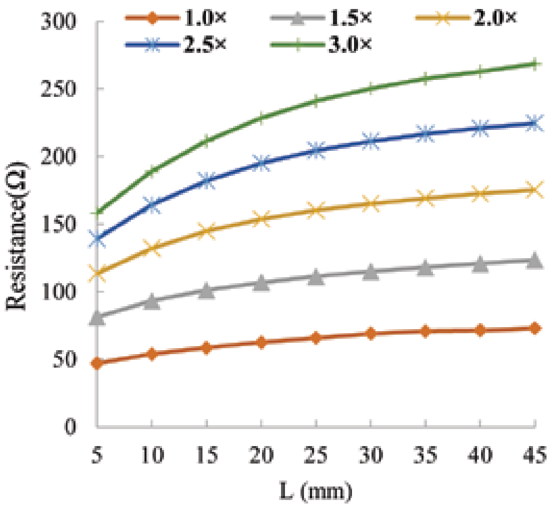



(a) (b)

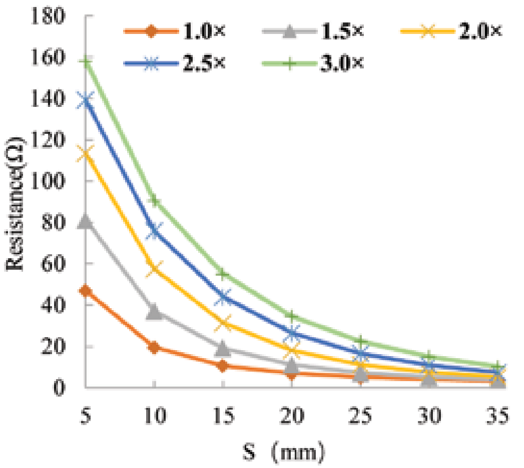
**Figure 10.** (a) The impedance selectivity of EIM variation with $S$. (b) The impedance selectivity of EIM variation with $L$, [3].

increased with increasing $S$ and $L$ and was not influenced by fat thickness. Furthermore, $Sel_m$ increased much faster with increasing $S$ compared to $L$. Taking the fat thickness of 1.5x as an example, for spacings $L$ = 5 mm and $S$ = 35 mm, $Sel_m$ was 90.95%, while at $S$ = 5 mm and $L$ = 35 mm, Selm was 32.34%.

The variation of EIM resistance with $S$ and $L$ and different fat thicknesses (between 6 mm, 1.0x and 18 mm, 3.0x) is shown in Fig. 11.




(a)




(b)

**Figure 11.** (a) The resistance stability of EIM variation with $L$. (b) The resistance stability of EIM variation with $S$, [3].

It was shown that at $L$ = 5 mm and $S$ = 5 mm (Fig. 11 a), the increase of resistance was 236.6% when the fat thickness increased from 1.0x to 3.0x. The increase of resistance was 423.3% with increasing fat thickness from 1.0x to 3.0x when $L$ = 45 mm. The resistance increased with increasing $L$ and fat thickness, indicating that the resistance stability decreased with increasing $L$. At $S$ = 5 mm and $L$ = 5 mm (Fig. 11 b), the resistance increased 236.66% as the fat thickness increased from 1.0x to 3.0x. The increase of resistance was 418.91% with increasing fat thickness from 1.0x to 3.0x when $S$ = 15 mm and $L$ = 5 mm. Although the resistance change showed a decreasing trend with increasing $S$, the resistance value decreased rapidly for smaller $S$ and slowly for larger values of $S$.

Considering requirements of the EIM electrode design for the parallel electrode configuration, such as the overall area of the electrodes, impedance amplitude, stability, and selectivity $Sel_m$ (muscle tissue contribution), distances $S$ = 10 mm and $L$ = 20 mm were finally chosen for the *in vivo* experimental study of local muscle fatigue. The EIM experimental platform is presented in Fig. 12. The EIM electrodes were placed in the middle of a *biceps brachii* muscle and connected to the Imp™ SFB7 anthropometric analyser working in the impedance spectrum mode. The angle between the upper and lower arm was constantly monitored using an angle sensor and the EMGworks software, in order to avoid changes in muscle length caused by inadvertent angle change during the experiments. Volunteers held dumbbells of different weights (0.1, 0.3, and 0.5 of their MVC) in their right arm. In the static experiment the angle between the lower arm and the body was $\theta = 45°$, while in the dynamic experiment it changed between $\theta = 0°$ and $\theta = 135°$, Fig. 13.


**Figure 12.** EIM experimental platform, [3].

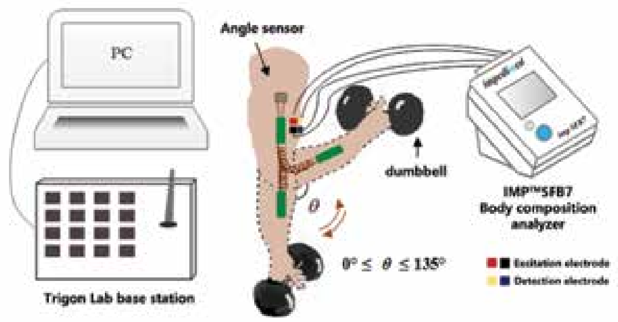



**Figure 13.** Block diagram of dynamic experiment, [3].

The trend of change in resistance during static contractions is shown in Fig. 8. $T_s$ was the time needed

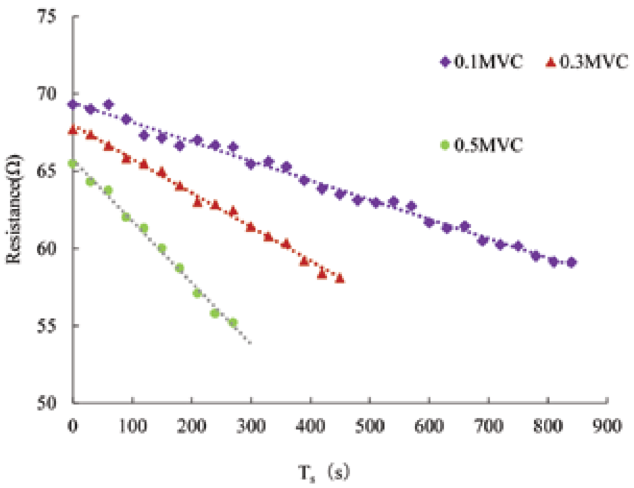



**Figure 14.** Static contraction experiments of muscles under different load conditions (0.1, 0.3, and 0.5 MVC). The decrease of $R$ (50 kHz) measured on Volunteer #1 from complete relaxation to extreme fatigue, [3].

for the muscle to change from a resting state to a fully fatigued state. When the muscle was in a state of complete fatigue, the EIM impedance remained constant. At $f$ = 50 kHz the EIM impedance value had a linear correlation with the degree of muscle fatigue, while the resistance had a negative correlation with the degree of fatigue, as in [1]. In addition, fatigue accumulation occurred more rapidly at higher MVCs, resulting in a higher slope of the linear curve and a shorter duration of the experiment $T_s$. For example, $T_s$ = 840 s under 0.1 MVC load, and $T_s$ = 270 s under 0.5 MVC load. To avoid individual differences, static experiments were performed on 10 volunteers. Although the initial EIM impedance values showed large differences, since they depend on the fat level of the volunteers, the resistance reduction between relaxed and exhausted muscle was about 10 Ω.

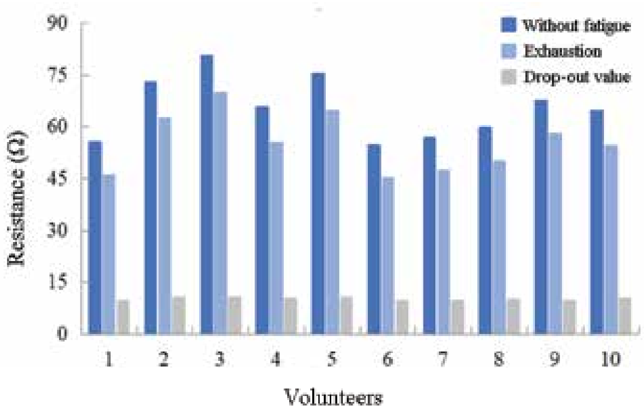



**Figure 15.** Resistance changes of 10 volunteers before (without fatigue) and after (exhaustion) the dynamic experiment for 0.3 MVC load, [3].

During dynamic contraction experiments the measured electrical impedance had a linear relationship with the number of contraction cycles, i.e. the duration of the experiment, for all three loads, which is consistent with the static experiment results. For heavier loads the experiments lasted shorter, indicating that the volunteer's fatigue progress was faster. The *biceps brachii* reached a completely fatigued state when the change in the amplitude of the EIM impedance value between two contraction cycles was extremely small and almost constant. EIM resistance $R$ measured on ten volunteers before and after the dynamic contraction experiments for 0.3 MVC load are presented in Fig. 15. A change of resistance between the beginning and the end of the exercise (exhaustion) varied around 10 Ω for all volunteers.

One of the first applications of a newly developed muscle fatigue monitoring technique was published in [5]. Neuromuscular electrical stimulation (NMES) is a common method for rehabilitation treatment and sports training, in which low-amplitude current is applied to the muscle tissue. However, a common side effect of the stimulation is muscle fatigue. The research in [5] focused on the changes in EIM parameters (impedance amplitude $|Z|$ and phase $\theta$) under muscle fatigue induced by the NMES. The effects of different stimulation parameters (amplitude, frequency, and pulse width) on NMES muscle fatigue were evaluated and analysed. The EIM parameters were compared to the mean power frequency (MPF) of sEMG thus verifying the validity of using EIM for evaluation of the muscle fatigue induced by NMES.

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