DualEMS: Two-Channel Arbitrary Waveform Electrical Muscle Stimulation Device to Design Interference Stimulation

Hiroki Ohara
Tokyo Institute of Technology
Japan
ohara.h.af@m.titech.ac.jp

Shoichi Hasegawa
Tokyo Institute of Technology
Japan
hase@pi.titech.ac.jp

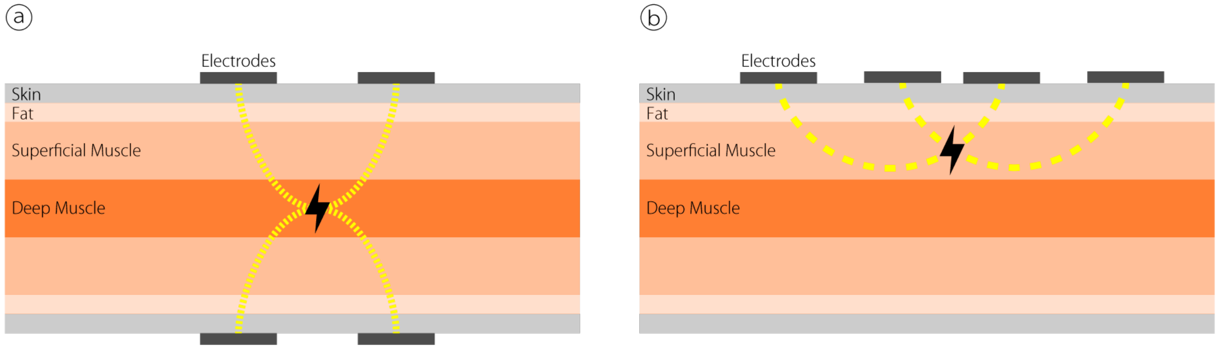

Figure 1: DualEMS is a haptic device engineered to design interference stimulation of electrical muscle stimulation (EMS). It achieves this by outputting multiple stimulus pulse of arbitrary waveforms simultaneously. Degrees of freedom of output waveform allow us to tackle two problems facing the current EMS device: (1) Difficulty in stimulating deep muscles; and (2) Unpleasant pain specific to electrical stimulation. Its high degree of freedom in interference conditions has made it possible to reproduce movements that have been difficult to achieve with conventional EMS. It thereby enables new applications that are not possible with existing EMS-based interactive devices. For instance, (a): We demonstrate EMS to deep muscle, that stimulates the pronatore teres muscle and supinator muscles and allows the forearm to rotate. (b): There were also unwanted movements driven by muscles that were not deep muscles, suggesting that interference stimulation can be applied to superficial muscles as well.

ABSTRACT

We propose a device to design interference stimulation, which has not been feasible in HCI. Also, we suggest parameters of stimulus pulses and electrodes' placement that facilitate stimulation of the supinator muscle (deep muscle). EMS (Electrical Muscle Stimulation) has been used in the medical field for a long time and, more recently, has been applied to HCI. It has been noticed that the diversity of stimulus pulses enables new stimulation methods. However, there is no compact EMS device that can output arbitrary waveforms in multichannel. Therefore, we developed "DualEMS", which can simultaneously output arbitrary waveforms with two channels. The user study validated the performance of the DualEMS, as it was possible to achieve the forearm pronation with the previously proposed method and the forearm supination with the newly proposed approach. In the future, we expect to realize new stimulation methods and interactions by using this system and approach.

CCS CONCEPTS

• **Human-centered computing → Haptic devices**; *HCI theory, concepts and models*; Gestural input.

KEYWORDS

Electrical deep muscle stimulation, Electrical muscle stimulation, Interference stimulation, Forearm pronation, Forearm supination

ACM Reference Format:
Hiroki Ohara and Shoichi Hasegawa. 2022. DualEMS: Two-Channel Arbitrary Waveform Electrical Muscle Stimulation Device to Design Interference Stimulation. In *Augmented Humans 2022 (AHs 2022), March 13–15, 2022, Kashiwa, Chiba, Japan.* ACM, New York, NY, USA, 8 pages. https://doi.org/10.1145/3519391.3519415

1 INTRODUCTION

What we wanted to achieve, in the first place, was to control the forearm rotation using EMS with surface electrodes, which previously had been uncertain. Electrical stimulation of the muscles that move the fingers and wrist can be used to control posture

Permission to make digital or hard copies of all or part of this work for personal or classroom use is granted without fee provided that copies are not made or distributed for profit or commercial advantage and that copies bear this notice and the full citation on the first page. Copyrights for components of this work owned by others than ACM must be honored. Abstracting with credit is permitted. To copy otherwise, or republish, to post on servers or to redistribute to lists, requires prior specific permission and/or a fee. Request permissions from permissions@acm.org.

AHs 2022, March 13–15, 2022, Kashiwa, Chiba, Japan

© 2022 Association for Computing Machinery.

ACM ISBN 978-1-4503-9632-5/22/03…$15.00
https://doi.org/10.1145/3519391.3519415

[31, 38, 39], present a sense of force [19, 20], and restore nerve function [21, 36]. However, it is still impossible to control the contraction of all muscles [5]. For example, when controlling the movement of the fingers with EMS, it is not feasible to control the flexor muscles of the thumb [38], and the pronator teres muscle, or the supinator of the forearm [35]. The wrist has three degrees of freedom, and EMS can control two of them (extension, flexion, ulnar deviation, and radial deviation) but not the other one (supination and pronation) [25]. All the muscles with the function of realizing the above movements are located in the deep layers [8, 33]. It is easy to stimulate deep muscles with implanted/percutaneous electrodes [22, 32], but we deliberated whether it was possible to do so with surface electrodes.

We considerd that the interference current stimulation (ICS) could be applied to induce muscle contraction in deep muscles (Figure 1). Interference is often involved in the stimulation of deep muscles using surface electrodes. It is believed that this method can achieve deep muscle contraction using surface electrodes, but there are three main concerns:

- Some studies have shown that the ICS can stimulate deep muscles, but the extent to which it induces muscle contraction has not been confirmed.
- In all studies, it is common to apply at least two stimulus pulses simultaneously to create an interference point, but the interference conditions (i.e., beat frequency) that cause muscle contraction are unknown.
- There are muscles for which the effectiveness of ICS has not yet been confirmed, such as the deep muscles of the forearm.

If these points are dispelled, we believe that the range of applications for EMS using surface electrodes will expand. In order to do this, the interference conditions need to be clarified, and a solution to these problems has also been proposed under certain conditions [26].

In order to design interference conditions of EMS, new EMS devices are required. Stimulators that generate stimulus pulses by switching can only generate rectangular stimulus pulses [15]. Several studies have investigated the effects of different stimulus waveforms on electrical stimulation [5, 29]. Therefore, in the field of HCI, there is a work that proposes a device that generates stimulation pulses with various waveforms in a single channel [16]. However, there is no compact EMS device capable of delivering multichannel (at least two) analog waveform stimulation pulses.

We have developed a device that can output various waveforms of electrical stimulation simultaneously in two channels. We termed our system "DualEMS", and the system is shown in Figure 2. We believe that this will enable us, using stimulation pulses with different waveforms, to stimulate multiple muscles simultaneously or to provide multiple electrical stimulations to a single muscle, which previously was difficult.

2 RELATED WORK

This section presents related work on electrical muscle stimulation for deep muscles, electrical stimulation devices and their safety, and related work dealing with forearm control.

2.1 Electrical Muscle Stimulation for Deep Muscle

Several studies have used implantable or percutaneous electrodes for the electrical stimulation of deep muscles [2, 22, 32, 37]. There are three types of electrodes used to apply electrical stimulation to muscle fibers: implanted, percutaneous, and surface. Implanted and percutaneous electrodes are mainly used for rehabilitation or to assist neural circuits. It can be reliably stimulated by targeting the nerves and muscle fibers, even those in the deeper layers. However, there is a concern that it is highly invasive, as the electrodes are inserted into the body. As far as implanted electrodes are concerned, they have become increasingly popular in recent years, including in pacemakers, due to the miniaturization of devices. Surface electrodes are also used in the field of rehabilitation, especially for analgesia and EMS. The electrodes are made of gel or cloth and are in direct contact with the skin, making them less invasive. However, it is less reliable as it stimulates the nerves and muscle fibers through the skin and fat. Furthermore, it is almost impossible to stimulate the deeper layers.

Therefore, two approaches have been considered for stimulating the deeper layers using surface electrodes. The first is focused on the parameters of the stimulation pulse and the electrical properties of the skin. Several studies have highlighted methods for the application of electrical stimulation deep in the human body through varying the parameters of the stimulus pulse. Furthermore, previous studies revealed that four parameters of electrical stimulation are essential to change the stimulation depth [3]. These are "waveform", "frequency", "amplitude" and "pulse width" [5, 29] (except in special cases, such as "Russian Current" [40], etc.). Of these factors, the waveform and frequency are essential to determine the depth of the stimulation. Previous studies have suggested that square waves do not penetrate deep muscles well, whereas sine waves pass through adipose tissue and reach deep muscles more efficiently [28, 30]. Regarding the relationship between frequency and skin resistance, Rosell et al. highlighted that the higher the frequency, the lower the resistance of the skin [34]. However, we generally uses low frequency (less than 1,000 Hz) stimulation pulses for EMS, and the effects of mid-frequency and high-frequency (over 1,000 Hz) pulses are unclear [5]. The second is to interfere with multiple stimulus pulses and control their by-products (interference points). It is possible to control the interference point to a high degree by increasing the number of electrodes [4], but the simplest method is the interferential current stimulation. Interferential current stimulation (ICS) is a method used for the electrical stimulation of deep muscles. ICS originated from "interferential therapy,"[1] which was used by Dr. Hans Nemec in the 1950s for cosmetic, rehabilitation, and pain-relief purposes [9]. Some of the conditions necessary for ICS to cause deep muscle contraction have been proposed [1, 10, 12, 26].

2.2 Electrical Muscle Stimulation Device

In the field of human-computer interaction (HCI), electrical stimulation is attracting attention as a promising technology. Originally, electrical stimulation was used for therapeutic purposes, and there are a number of electrical stimulators available as medical

[1] http://www.hans-nemec.at/interferenz.php

devices.[2][3] Due to its characteristics to move the body by controlling the contraction of one's own muscles, EMS is also used in the field of rehabilitation [27] and art [7]. Researchers in the field of HCI therefore focused on two things: the compactness of the device and its ability to present a sense of force by treating the user's muscles as actuators. This is why many researchers have used EMS to study HCI and VR in the past decade [17].

Researchers have developed tools for electrical stimulation for HCI, but the performance of the devices is still limited. Stimulators used in therapy and rehabilitation have a variety of parameters of stimulus pulse, but are too large for use in HCI. Therefore, compact stimulators have been proposed in the field of HCI. In some studies, miniaturization of the stimulator was achieved by boosting the signal from the microcontroller and controlling the frequency and pulse width of the stimulation pulses with a switch such as a photo-MOS relay [15, 24, 31, 39]. There have also been several proposals for multichannel electrical stimulators [6, 13, 15, 31]. However, the ability to output stimulation pulses with multiple analog waveforms has not yet been proposed, and we have developed an electrical stimulator that provides this functionality on a single circuit board.

3 DUALEMS: TWO-CHANNEL ARBITRARY WAVEFORM ELECTRICAL MUSCLE STIMULATOR

This section describes the EMS equipment developed to realize the EdMS.

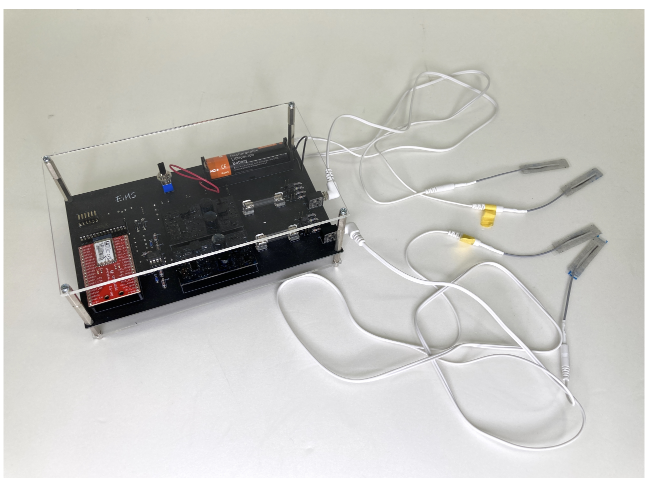

Figure 2: DualEMS.

3.1 System Overview

An overview of the system is given in Figure 3. The output channels and the parameters of the stimulation pulses (waveform, pulse width, monophasic/biphasic, amplitude inversion, each frequency, and stimulation time) can be controlled by the GUI of our original software. The stimulation pulses are sent as audio data to a Bluetooth module on the board. The signal sent is adjusted to 0 V by an operational amplifier, amplified through the amplifier module, and sent to the electrode. A single board can output up to two independent stimulation pulses, which can easily be increased by adding more Bluetooth connections. However, the same stimulation pulse is output from the same channel on each board.

In designing the system, the following features were considered:

- The number of parameters and stimulation pulses needed to achieve interference should be easily available. In other words, the device should be able to output several stimulation pulses of analog waveform such as sin wave at the same time.
- User safety must be ensured as far as possible. This means two things: ensuring that the EMS device is physically isolated within the system, and ensuring that the safety standards for electrical stimulation are not exceeded.

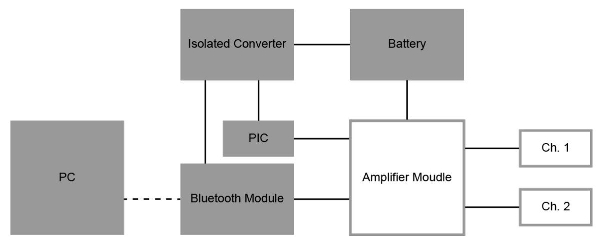

Figure 3: System overview of the DualEMS.

3.2 Implementation

Hardware. We designed an original electric circuit and newly made circuit boards (Figure 4). The hardware mainly consists of a Bluetooth module (RN-52, Microchip Technology Inc.), two PWM amplifiers (IFJM-001, Marutsuelec Co., Ltd.), two fuses (0312.062, Littelfuse, Inc.), a microcontroller, and a battery (Figure 3). In order to prevent the PWM amplifier from outputting before the Bluetooth connection, the microcontroller controls the PWM amplifier's CSD pin ON/OFF. The fuses are of the physically blown type and are replaceable.

In designing the hardware, we took care of the safety in two respects: that it is electrically isolated and that there is no overcurrent. Firstly, we have achieved electrical isolation by using wireless communication between the EMS device and the PC. With reference to [16], only the power supply of the PWM amplifier was connected directly to the power supply of the whole device, while the other components were isolated from the power supply of the whole device. This eliminates the risk of current flowing directly from the battery to the electrodes. If the stimulus pulses are generated directly from the microcontroller, the output part of the stimulus pulse should be galvanically isolated [15, 24]. With reference to IEC 60601-2-10:2012 and JIS T 0601-2-10:2015, we have selected a fuse with a voltage rating of 62 mA, which is the lowest on the market. In addition, [18] has stricter limits on the contact current.

[2]https://www.healthcare.omron.co.jp/product/hvf/
[3]https://www.medical.itolator.co.jp/product/

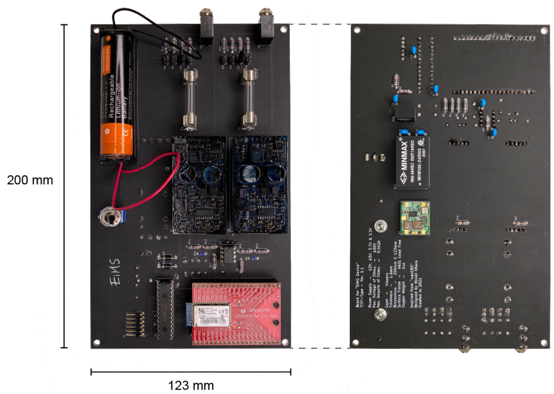

Figure 4: Hardware implementation.

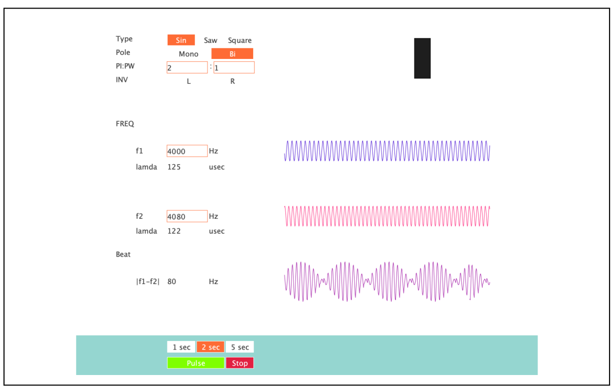

Figure 5: GUI of the software.

Software. The software and GUI were implemented using Processing (3.5.4) (Figure 5). The library "minim" is used to send the generated audio data as stimulus pulses to the Bluetooth module. To determine the amplitude of the stimulation pulse, the volume of the transmitted signal is adjusted with the volume of the PC. For each stimulus, the parameters of the stimulus pulse are logged in a cvs file, which can later be used for statistical analysis of the results.

In designing the software, we paid attention to two points: the stimulus pulses must not generate unexpected parameters, and the software must be able to communicate with external force sensors for the future experiment. The first is a software safety measure. From a safety point of view, it is not a good idea to apply a direct current to the body. Three restrictions are therefore imposed: all parameters must be set, the frequency must be above 20 Hz, and the pulse width must not exceed the pulse frequency. If the restriction conditions are not met, the user is not allowed to press the button "Pulse." Secondly, we have ensured the scalability of the system. In the research of the interaction by the electric stimulation, there is a method to judge the state of the body before and after applying the electric stimulation by the angle (image data). However, in order to compare the changes between MVC or MVIC and EMS stimulation, the advantage of being able to acquire time series data of force sensation in six axes at the same time as stimulation is significant. In addition, the values read from the sensors can be displayed in the GUI in real time.

3.3 Performance

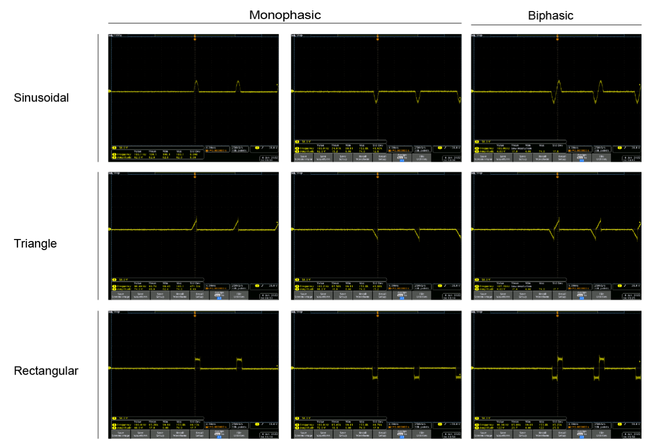

Figure 6: Each stimulus pulses of the DualEMS.

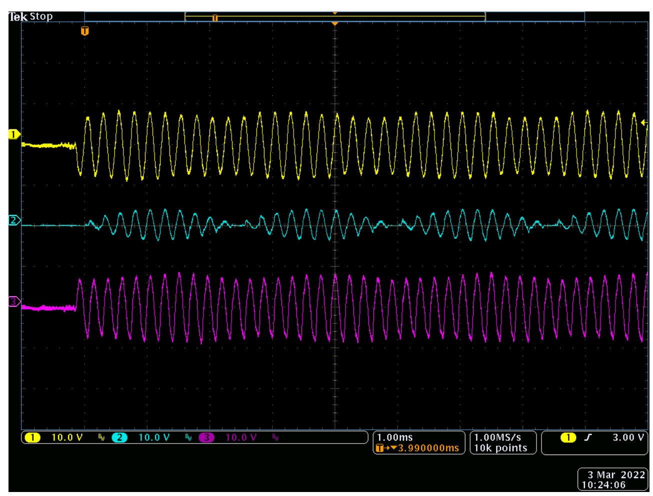

Figure 7: The yellow waveform represents stimulus pulses at 4,000 Hz and the pink waveform represents stimulus pulses at 4,080 Hz. The light blue waveform represents the stimulus pulse at the interference point. Four couples of paralleled 4.7 kΩ resistor and 0.01 μF capacitor are laid out in star-shaped pattern for the measurements.

We measured the performance results of the Dual EMS using the software described above. The frequency was set at 100 Hz, and the pulse frequency and pulse width at a duty ratio of 8:1. As a result, monophasic and biphasic outputs were successfully achieved for all waveforms (Figure 6). It was also possible to observe that muscle contraction occurs using the waveform (frequency: 80 Hz, pulse width: 390 μsec, waveform: rectangular, phase: biphasic). We adjusted the value of gain for each waveform because different waveforms have different stimulus intensities, even if the audio output of the PC that generates the signal is the same.

We also confirmed that two stimulus pulses can be output simultaneously and that each parameter can be set independently, except for the stimulation time (Figure 7). There is a fine noise in the waveform, which we believe is due to the fact that the IFJM-001 is controlled by PWM in particular. It can also be connected to other DualEMS via Bluetooth to further increase the number of stimulation pulses. However, because the stimulus pulses are generated from the PC's audio signal, the maximum number of types of stimulus pulses that can be output simultaneously is two.

4 USER STUDY: DRIVE FOREARM ROTATION USING DUALEMS

In this section, we challenge whether it is feasible for not only forearm pronation but also forearm supination using the DualEMS.

4.1 Overview

To date, the method of achieving forearm pronation with EMS using surface electrodes is ambiguous [23, 35] and there is no precedent for stimulating the pronator teres muscle and the supinator muscle with ICS. The pronator teres muscle and the supinator muscle are subject to the following conditions:

- Function of the target deep muscle, which is covered by several muscles
- Body movements that can only be realized by the target deep muscle and not by the covered muscles

In the user study, we investigated whether the forearm pronation and supination could be induced by the EdMS using the DualEMS. EdMS, which is proposed by Ohara et al., is a method of creating low-frequency electrical stimulation in the body by interfering with stimulation pulses of a higher frequency than that of normal EMS [26].

4.2 Participants

Three healthy individuals (three males) participated in the user study (Table 1). One was unaccustomed to electrical stimulation, and two were accustomed to electrical stimulation. All participants were recruited from our laboratory. The Human Subject Research Ethics Review Committee at Tokyo Institute of Technology approved the study protocol, and the approval number is 2020255. Written informed consent was obtained from each participant prior to the initiation of the user study. No adverse events were associated with this study.

4.3 Procedure

The user study was conducted in two stages: calibration and investigation. The calibration determined the placement of the electrodes and the strength of the peak amplitude. Regardless of which hand is dominant, the right forearm is farther from the heart than the left forearm; thus, the right forearm of each participant was electrically stimulated during the user study. Each participant placed his or her right forearm on the desk and four electrodes (two pairs) were placed so that the proposed method (EdMS) would be most effective. We were careful to note that as the joints moved, the relative positions of the muscles and skin changed [11].

After the calibration was complete, electrical stimulation was given for one second and measured three times in each condition. A camera was placed in front of the forearm of the participant to capture the image of the forearm rotating. The results of rotation angle was calculated from the difference between the zero position and the maximum rotation position of the line segment connecting the MP joints of the index and middle fingers on the video. The timing of the stimulation was informed by voice so that the subject would not be startled and make unnecessary movements.

4.4 Parameters of the Stimulus Pulse

In the user study, the parameters were set as follows:

- The waveform and the pulse width are sinusoidal ("pulse frequency" : "pulse width" = 2:1).This was chosen because a simple sine wave is said to reach the deeper layers more easily than other waveforms [28, 30], perhaps because there are fewer low-frequency components than, for example, a square wave.
- The amplitude is set for each user and is one level lower than the level of pain-like distress on the skin.
- The pulse frequency was set at 4,000 Hz for one stimulus pulse and 4,080 Hz for the other. This produces a low-frequency (80 Hz) stimulation pulse at the interference point.

4.5 Apparatuses

We used the DualEMS for the user study. Regarding the electrodes, we used four gel electrodes (Axelgaard Manufacturing Co., Ltd.). Gel electrodes are considered to be the least painful among the surface electrodes [14]. The size of each electrode was 50 mm × 50 mm.

4.6 Placement of the Electrodes

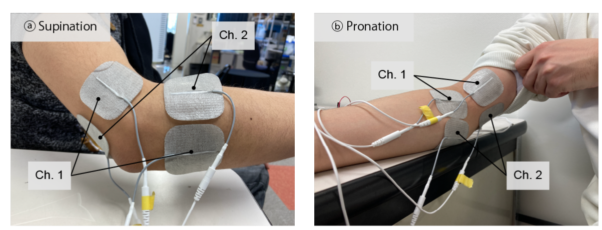

Figure 8: Placement of electrodes.

We proposed an electrode placement to achieve the forearm supination. As shown in Figure 9, two sets of gel electrodes are used. A pair of gel electrodes, an anode and a cathode, is placed along the direction of the muscle fiber. The two pairs of gel electrodes should be placed in parallel or intersecting directions so that the same poles are close together. Arrangements had been proposed to achieve forearm pronation by interference, but forearm supination had not yet been achieved [26].

Table 1: Participants' Characteristics.

Participant	Gender	Age	Forearm Circumference [cm]	Peak Amplitude [V]
00	Male	23	24.5	(Pro.) 19, (Supi.) 34
01	Male	24	23	(Pro.) 19, (Supi.) 26
02	Male	30	26	(Pro.) 34, (Supi.) 45

In the EdMS, stimulation pulses of different pulse frequencies are output from each electrode pair to generate interference waves. Preliminary experiments show that there is no dependence between electrode pairs and interference conditions. This means that if one wishes to produce a beat frequency of 80 Hz with reference frequencies of 4000 Hz and 4080 Hz, one does not need to worry about which electrode pair produces 4000 Hz.

4.7 Results

In the user study, all participants were able to achieve the forearm pronation and supination as shown in Figure 8. Firstly, overall, pronation averaged 66 degrees (SD = 16.09, n = 3) and supination averaged 48.7 degrees (SD = 6.03, n = 3).

The results for each participant individually were as follows: the mean pronation of participant 01 was 45 degrees (SD = 7.94, n = 3), and the mean supination was 53.7 degrees (SD = 3.22, n = 3). The mean pronation of participant 02 was 76.3 degrees (SD = 6.43, n = 3), and the mean supination was 48 degrees (SD = 1, n = 3). The mean pronation of participant 03 was 72 degrees (SD = 6.56, n = 3), and the mean supination was 40.7 degrees (SD = 4.93, n = 3).

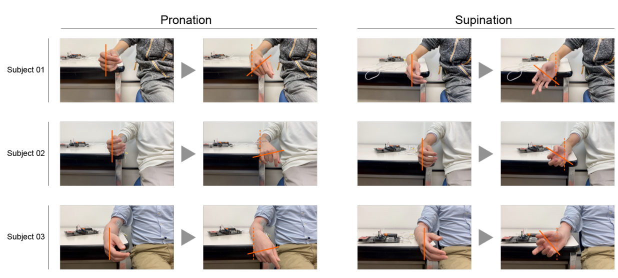

Figure 9: Results of each participant.

5 DISCUSSION

During the development of the EMS equipment, great care was taken not only to achieve the required performance but also to prevent unexpected electric shocks. The DualEMS is controlled by a PIC to ensure that no noise is output from the PWM Amplifier before the Bluetooth connection is made. In addition, fuses and resistors are placed between the PWM Amplifier and the electrodes to prevent dangerous currents from flowing. Furthermore, because the DualEMS communicates wirelessly with the signal output device (PC), it also solves the problem of galvanic isolation. As there is no galvanic isolation between the PWM amplifier and the battery, an acrylic board prevents the user from touching the board directly.

In the user study, forearm pronation and supination were achieved by using a higher pulse frequency than that used in common EMS. However, there were some unwanted movements. It was a movement with superficial muscles, which suggests that depending on the position and size of the electrodes, not only deep but also superficial muscles can be stimulated. The higher the frequency, the harder it is for humans to perceive the electrical stimulus. Therefore, the pain (discomfort) in the skin that has been felt with the use of EMS may be eliminated with the use of this device.

Then, what we aim to do in the future is as follows:

- Measure the time resolution of the device to investigate the time response of the EMS/EdMS
- Further miniaturization of the substrate to make it more wearable

5.1 Findings

In our user study, we got three findings. The first was that the unwanted movements were common to all participants. The unnecessary movements were, in the case of supination, the movement of the middle finger extending and the movement of the thumb extending. In the case of pronation, it was the flexion of the middle, ring, and little fingers. However, with the pulse frequency used in this study, the size of the electrodes can be further reduced, and the stimulation area can be narrowed. Secondly, when we were discussing the application with the participants after the user study, many of them commented that it was easier to be rotated if the hand was in the air. In the user study, it took some time to calibrate the electrodes' placement with a hand on the desk. This is probably because the gel electrode is sandwiched between the desk and the arm, and the relative distance between the electrode position and the muscle is different. Thirdly, pronation might be easier to realize than spination. This thought is based on the difference in the magnitude of the amplitude required for realization and the average size of the rotation angle. This makes sense, given that anatomically the pronator teres muscle is more superficial than the supinator muscle.

5.2 Limitations

There are two limiting factors of our device. One is that the number of stimulus pulses that can be output simultaneously is limited to two. By increasing the number of Bluetooth connections, it is possible to output stimulation pulses from multiple DualEMS simultaneously. However, since the signal source is audio data from the PC, there is a limit to the number of types of stimulation pulses that can be produced simultaneously. The other limitation is that our device can only modulate the voltage of the stimulus pulse. In the ICS, as the name suggests, current modulation is the most common method. The EdMS is a voltage modulated version of this technique, which we have been able to realise it using DualEMS. Even at the same voltage, if the amount of current is different, the perception of the stimulus and the intensity of the muscle contraction will be different. Therefore, if the modulation of the current could also be controlled, more detailed manipulation would be possible.

Also, the small number of people in the user study is a limitation. We have shown that the performance of the proposed device is sufficient and that "EdMS" is likely to be effective. However, due to the small number of participants (three) and the fact that all of them were men, we could not have an in-depth discussion this time. Also, the amplitude of the stimulation pulse was set according to the pain, so technically it should be possible to obtain a larger rotation angle.

6 CONCLUSION

This paper introduced the DualEMS, a two-channel arbitrary waveform EMS device, and a method to stimulate the supinator muscles. In the field of HCI, a number of EMS-based applications have been proposed in the last decade, but few of them refer to the design of stimulation pulses. Recently there has been a growing interest in EMS using high-frequency electrical stimulation to reduce skin pain, but high-frequency electrical stimulation is not suitable for EMS. However, we thought it would make sense to propose a device that could be used in the field of HCI, where methods using high frequency electrical stimulation are being explored. Thus, we have created the DualEMS, its software, and GUI. It also showed the practicality of the device and supported the effectiveness of one EMS method, "EdMS". In this study, the stimulation pulses and interference conditions were set up in such a way as to target deep muscles. In the future, we hope to be able to design different stimulation pulses and interference conditions to reduce pain, reduce the size of the electrodes, and refine the stimulation of superficial muscles and deep muslces.

REFERENCES

[1] S.D. Bennie, J.S. Petrofsky, J. Nisperos, M Tsurudome, and M Laymon. 2002. Toward the optimal waveform for electrical stimulation of human muscle. *Eur J Appl Physiol* 88 (2002), 13–19. https://doi.org/10.1007/s00421-002-0711-4

[2] N Bhadra, KL Kilgore, and PH Peckham. 2001. Implanted stimulators for restoration of function in spinal cord injury. *Medical engineering & physics* 23, 1 (2001), 19–28.

[3] C.S. Bickel, C.M. Gregory, and J.C. Dean. 2011. Motor unit recruitment during neuromuscular electrical stimulation: a critical appraisal. *Eur J Appl Physiol* 111 (2011). Issue 2399. https://doi.org/10.1007/s00421-011-2128-4

[4] S. Bounyong, S. Adachi, T. Yoshimoto, T. Ota, and J. Ozawa. 2016. Controlling interfered area in interferential current stimulation by electrode-area patterning. In *2016 38th Annual International Conference of the IEEE Engineering in Medicine and Biology Society (EMBC)*. 1721–1724. https://doi.org/10.1109/EMBC.2016.7591048

[5] B M. Doucet, A. Lam, and L. Griffinb. 2012. Neuromuscular electrical stimulation for skeletal muscle function. *Yale J Biol Med* 85(2) (2012), 201–215.

[6] Tim Duente, Max Pfeiffer, and Michael Rohs. 2017. Zap++: A 20-Channel Electrical Muscle Stimulation System for Fine-Grained Wearable Force Feedback. In *Proceedings of the 19th International Conference on Human-Computer Interaction with Mobile Devices and Services (MobileHCI '17)*. Association for Computing Machinery, Article 1, 13 pages. https://doi.org/10.1145/3098279.3098546

[7] Arthur Elsenaar and Remko Scha. 2002. Electric Body Manipulation as Performance Art: A Historical Perspective. *Leonardo Music Journal* 12 (2002). https://doi.org/10.1162/096112102762295089

[8] F.H.Netter. 2014. *Atlas of Human Anatomy (6th edition)*.

[9] J.M. Ganne. 1975. INTERFERENTIAL THERAPY. *J Physiother* 22 (1975), 101–110. https://doi.org/10.1016/S0004-9514(14)61005-9

[10] Meltem MD Gundog, Funda MD Atamaz, Selcen MD Kanyilmaz, Yesim MD Kirazli, and Gunay MD Celepoglu. 2012. Interferential Current Therapy in Patients with Knee Osteoarthritis. *American Journal of Physical Medicine & Rehabilitation* 91 (2012), 107–113. Issue 2. https://doi.org/10.1097/phm.0b013e3182328687

[11] Kento Ichikawa, Yinlai Jiang, Masao Sugi, Shunta Togo, and Hiroshi Yokoi. 2021. Joint angle based motor point tracking stimulation for surface FES: A Study on biceps brachii. *Medical Engineering & Physics* 88 (2021), 9–18. https://doi.org/10.1016/j.medengphy.2020.11.014

[12] Fuentes Jorge P., Olivo Susan Armijo, Magee David J., and Gross Douglas P. 2010. Effectiveness of Interferential Current Therapy in the Management of Musculoskeletal Pain: A Systematic Review and Meta-Analysis. *Physical Therapy* 90 (2010), 1219–1238. Issue 9. https://doi.org/10.2522/ptj.20090335

[13] Hiroyuki Kajimoto. 2012. Electrotactile Display with Real-Time Impedance Feedback Using Pulse Width Modulation. *IEEE Transactions on Haptics* 5, 2 (2012), 184–188. https://doi.org/10.1109/TOH.2011.39

[14] Thierry Keller and Andreas Kuhn. 2008. Electrodes for transcutaneous (surface) electrical stimulation. *Journal of Automatic Control* 18 (2008), 35–45. Issue 2. https://doi.org/10.2298/JAC0802035K

[15] Michinari Kono, Yoshio Ishiguro, Takashi Miyaki, and Jun Rekimoto. 2018. Design and Study of a Multi-Channel Electrical Muscle Stimulation Toolkit for Human Augmentation. In *Proceedings of the 9th Augmented Human International Conference (AH '18)*. Association for Computing Machinery, Article 11, 8 pages. https://doi.org/10.1145/3174910.3174913

[16] Michinari Kono and Jun Rekimoto. 2019. wavEMS: Improving Signal Variation Freedom of Electrical Muscle Stimulation. In *2019 IEEE Conference on Virtual Reality and 3D User Interfaces (VR)*. 1529–1532. https://doi.org/10.1109/VR.2019.8798102

[17] Michinari Kono, Takumi Takahashi, Hiromi Nakamura, Takashi Miyaki, and Jun Rekimoto. 2018. Design Guideline for Developing Safe Systems That Apply Electricity to the Human Body. *ACM Trans. Comput.-Hum. Interact.* 25 (2018), 19:1–19:36. https://doi.org/10.1145/3184743

[18] Jiali Lin, R. Saunders, K. Schulmeister, P. Söderberg, A. Swerdlow, M. Taki, Bernard Veyret, G. Ziegelberger, Mike H. Repacholi, Rüdiger Matthes, et al. 2010. ICNIRP Guidelines for limiting exposure to time-varying electric and magnetic fields (1 Hz to 100 kHz). *Health physics* 99 (2010), 818–836.

[19] Pedro Lopes and Patrick Baudisch. 2013. Muscle-Propelled Force Feedback: Bringing Force Feedback to Mobile Devices. In *Proceedings of the SIGCHI Conference on Human Factors in Computing Systems* (Paris, France) *(CHI '13)*. Association for Computing Machinery, New York, NY, USA, 2577–2580. https://doi.org/10.1145/2470654.2481355

[20] Pedro Lopes, Sijing You, Alexandra Ion, and Patrick Baudisch. 2018. Adding Force Feedback to Mixed Reality Experiences and Games Using Electrical Muscle Stimulation. (2018), 1–13. https://doi.org/10.1145/3173574.3174020

[21] Nebojša M. Malešević, Lana Z. Popović Maneski, Vojin Ilić, Nikola Jorgovanović, Goran Bijelić, Thierry Keller, and Dejan B. Popović. 2012. A multi-pad electrode based functional electrical stimulation system for restoration of grasp. *Journal of neuroengineering and rehabilitation* 9, 1 (2012), 1–12.

[22] E. Byron Marsolais and Rudi Kobetic. 1987. Functional electrical stimulation for walking in paraplegia. *J Bone Joint Surg* 69, 5 (1987), 728–733.

[23] Arinobu Niijima, Toki Takeda, Ryosuke Aoki, and Yukio Koike. 2021. Reducing Muscle Activity When Playing Tremolo by Using Electrical Muscle Stimulation. In *Augmented Humans Conference 2021* (Rovaniemi, Finland) *(AHs'21)*. Association for Computing Machinery, New York, NY, USA, 289–291. https://doi.org/10.1145/3458709.3458977

[24] Jun Nishida and Kenji Suzuki. 2017. BioSync: A Paired Wearable Device for Blending Kinesthetic Experience. In *Proceedings of the 2017 CHI Conference on Human Factors in Computing Systems (CHI '17)*. Association for Computing Machinery, 3316–3327. https://doi.org/10.1145/3025453.3025829

[25] Cynthia C. Norkin and D. Joyce White. 2016. *Measurement of joint motion: a guide to goniometry.* FA Davis.

[26] Hiroki Ohara and Shoich Hasegawa. 2022. An Electrical Stimulation Method to Control Deep Muscle Contraction using Surface Electrodes. (2022). https://doi.org/10.21203/rs.3.rs-1201798/v1

[27] P. Hunter Peckham and Jayme S. Knutson. 2005. Functional Electrical Stimulation for Neuromuscular Applications. *Annual Review of Biomedical Engineering* (2005), 327–360. https://doi.org/10.1136/bjsm.24.2.87

[28] J. Petrofsky. 2008. The effect of the subcutaneous fat on the transfer of current through skin and into muscle. *Medical Engineering and Physics* 30 (2008), 1168–1176. https://doi.org/10.1016/j.medengphy.2008.02.009

[29] J. Petrofsky, M. Laymon, M. Prowse, S. Gunda, and J. Batt. 2009. The transfer of current through skin and muscle during electrical stimulation with sine, square, Russian and interferential waveforms. *Journal of Medical Engineering & Technology* 33 (2009), 170–181. Issue 2. https://doi.org/10.1080/03091900802054580

[30] J. Petrofsky, M. Prowse, M. Bain, E. Ebilane, H. J. Suh, J. Batt, D. Lawson, V. Hernandez, A. Abdo, T. Yang, E. Mendoza, K. Collins, and M. Laymon. 2008.

Estimation of the distribution of intramuscular current during electrical stimulation of the quadriceps muscle. *Eur J Appl Physiol* 103 (2008), 265–273. https://doi.org/10.1007/s00421-008-0700-3

[31] Max Pfeiffer, Tim Duente, and Michael Rohs. 2016. Let Your Body Move: A Prototyping Toolkit for Wearable Force Feedback with Electrical Muscle Stimulation. In *Proceedings of the 18th International Conference on Human-Computer Interaction with Mobile Devices and Services (MobileHCI '16)*. Association for Computing Machinery, 418–427. https://doi.org/10.1145/2935334.2935348

[32] K.T. Ragnarsson. 2008. Functional electrical stimulation after spinal cord injury: current use, therapeutic effects and future directions. *Spinal cord* 46, 4 (2008), 255–274.

[33] J.W. Rohen, C. Yokochi, and E. Lütjen-Drecoll. 2016. *Anatomy: A Photographic Atlas (8th edition)*.

[34] J. Rosell, J. Colominas, P. Riu, R. Pallas-Areny, and J.G. Webster. 1988. Skin impedance from 1 Hz to 1 MHz. *IEEE Transactions on Biomedical Engineering* 35 (1988), 649–651. https://doi.org/10.1109/10.4599

[35] Mose Sakashita, Yuta Sato, Ayaka Ebisu, Keisuke Kawahara, Satoshi Hashizume, Naoya Muramatsu, and Yoichi Ochiai. 2017. Haptic Marionette: Wrist Control Technology Combined with Electrical Muscle Stimulation and Hanger Reflex. In *SIGGRAPH Asia 2017 Posters* (Bangkok, Thailand) *(SA '17)*. Association for Computing Machinery, New York, NY, USA, Article 33, 2 pages. https://doi.org/10.1145/3145690.3145743

[36] A. Selfslagh, S. Shokur, D.S.F. Campos, A.R.C. Donati, S. Almeida, S.Y. Yamauti, D.B. Coelho, M. Bouri, and M.A.L. Nicolelis. 2019. Non-invasive, Brain-controlled Functional Electrical Stimulation for Locomotion Rehabilitation in Individuals with Paraplegia. *Nat Commun* 9 (2019), 6782. https://doi.org/10.1038/s41598-019-43041-9

[37] B. Smith, Zhengnian Tang, M.W. Johnson, S. Pourmehdi, M.M. Gazdik, J.R. Buckett, and P.H. Peckham. 1998. An externally powered, multichannel, implantable stimulator-telemeter for control of paralyzed muscle. *IEEE Transactions on Biomedical Engineering* 45, 4 (1998), 463–475. https://doi.org/10.1109/10.664202

[38] Akifumi Takahashi, Jas Brooks, Hiroyuki Kajimoto, and Pedro Lopes. 2021. *Increasing Electrical Muscle Stimulation's Dexterity by Means of Back of the Hand Actuation*. Association for Computing Machinery, New York, NY, USA. https://doi.org/10.1145/3411764.3445761

[39] Emi Tamaki, Takashi Miyaki, and Jun Rekimoto. 2011. PossessedHand: Techniques for Controlling Human Hands Using Electrical Muscles Stimuli. In *Proceedings of the SIGCHI Conference on Human Factors in Computing Systems (CHI '11)*. Association for Computing Machinery, 543–552. https://doi.org/10.1145/1978942.1979018

[40] A. Ward and N. Shkuratova. 2002. Russian Electrical Stimulation: The Early Experiments. *Physical Therapy* 82 (2002), 1019–1030. Issue 10. https://doi.org/10.1093/ptj/82.10.1019