

Zap++: A 20-Channel Electrical Muscle Stimulation System for Fine-Grained Wearable Force Feedback

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ABSTRACT

Electrical muscle stimulation (EMS) has been used successfully in HCI to generate force feedback and simple movements both in stationary and mobile settings. However, many natural limb movements require the coordinated actuation of multiple muscles. Off-the-shelf EMS devices are typically limited in their ability to generate fine-grained movements, because they only have a low number of channels and do not provide full control over the EMS parameters. More capable medical devices are not designed for mobile use or still have a lower number of channels and less control than is desirable for HCI research. In this paper we present the concept and a prototype of a 20-channel mobile EMS system that offers full control over the EMS parameters. We discuss the requirements of wearable multi-electrode EMS systems and present the design and technical evaluation of our prototype. We further outline several application scenarios and discuss safety and certification issues.

Author Keywords

Electrical muscle stimulation; electrode grid; wearable; mobile; wearable force feedback; mobile haptic output

ACM Classification Keywords

H.5.2. Information Interfaces and Presentation: User Interfaces – Haptic I/O, input devices and strategies

INTRODUCTION

Electrical muscle stimulation (EMS) is used in fitness training, wellness applications, medical applications, and increasingly also in HCI. Off-the-shelf devices that have been developed for medical applications, such as rehabilitation, are not ideally suited for HCI. The shape of the EMS signal cannot be adjusted so as to optimize the user experience and comfort of the signal, as this normally is not an issue with medical applications. The number of simultaneous EMS channels of such devices is typically limited to one to four, which is insufficient for many purposes in HCI. Moreover, many of the

medical devices are designed for stationary use and are not easy to deploy in mobile contexts. Some special medical devices exist that enable the real-time adjusting of stimulation parameters, but these are not mobile [19].

Natural movements require the coordinated actuation of multiple muscles in a precisely defined temporal pattern. To generate natural movements existing research in HCI has employed complex configurations of multiple independent signal-generating devices (see for example [9]). However, this workaround is bulky and not suitable for mobile applications. Mobile EMS systems have to be as compact and energy efficient as possible. Off-the-shelf devices are essentially black boxes and not flexible enough to be used as research platforms. The limits of off-the-shelf devices increasingly constrain further advancements of HCI research involving EMS. Hence there is a clear need for more advanced multi-electrode EMS devices that offer detailed control over the EMS signal parameters. The system design described in this paper is a first step towards addressing this need. However, important safety and device certification aspects remain to be solved before the proposed system may be widely used. We stress that this is a research prototype that we have not yet used in experiments with participants (except for ourselves).

In particular, we present the concept and a prototype implementation of a 20-channel wearable electrical muscle stimulation system that offers full control over the EMS signal parameters. Different electrode grids can be connected to the signal generator depending on the targeted muscles and the point of application on the human body (arm, leg, shoulder). Our current prototype uses a multi-electrode grid on the lower arm, consisting of two rings, similar to [32]. The prototype enables the fine-grained actuation of the hand and fingers. Parameters can be set individually for each electrode pair even during the stimulation. The detailed control of EMS signal parameters allows taking into account user experience aspects (e.g., comfort optimization [2]), performance aspects (e.g., reducing muscle fatigue through frequency change [13]), as well as creating various sensations with electrical stimulation of the skin (e.g., warmth, tickling, pressure [29], or itch [33]).

EMS has many applications in mobile contexts. Examples include pedestrian navigation (automatically steering pedestrians [23]), force-feedback for handheld games (device tilts depending on game state [16]), and pervasive exertion games (force feedback during jogging [8], feedback when hitting a ball [7] or an object [25]). Existing work focuses on effec-

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tive but simple force feedback, often activating only a single muscle. More challenging mobile augmented reality scenarios include augmenting objects in the real world through EMS feedback [18], haptically indicating points of interests to the user [14], or simulating virtual objects in the real world so that the user can feel them physically [27].

Augmented reality applications such as PokémonGo¹, Anatomy 4D², or Aurasma³, and smart glasses such as HoloLens⁴ or Moverio BT-200⁵ are on the rise. However, these technologies mainly stimulate the visual sense. In these cases the simulated objects do not have any physical properties such as weight or stiffness. For example, grasping, throwing, and catching Pokéballs would be much more entertaining if they had physical properties like mass, inertia, and surface stiffness as in *GyroTab* [1] or *muscle-plotter* [19]. Creating realistic proprioceptive feedback even for a simple virtual ball is challenging, because it involves almost all fingers and the connected muscles. Augmented reality offers the opportunity to extend virtual and physical objects by physical properties [7]. For example a virtual ball might have more simulated weight over stone floors or it may start to shake over grass land to indicate that it should be thrown (cf. [18]).

Adding virtual haptic properties to objects using EMS involves actuating many muscles of the hand, the lower and the upper arm. Multiple muscles need to be actuated at the same time and some muscles need to produce a counter force for other muscles. The described 20-channel mobile electrical muscle stimulation system can be a building block for this kind of applications, but has to be combined with suitable sensory systems.

The rest of this paper is structured as follows. The next sections outline a number of application scenarios that the presented system supports and give an overview of related work in the area, respectively. Then we discuss the requirements of wearable multi-electrode EMS systems for use in mobile HCI. The core of this paper presents the details of our hardware prototype and its technical evaluation. We describe the calibration process in a separate section as it is challenging for multi-electrode systems. The final section provides important considerations in terms of device safety and certification.

APPLICATION SCENARIOS

Haptic output is difficult to realize in mobile scenarios, especially when the goal is to move body parts. The required forces are typically generated by motors or other actuators, which consume significant amounts of power and are bulky in mobile use, particularly if attached to the arm, hands, or fingers. Thus, it is still challenging to generate high fidelity force feedback in a mobile context. EMS technology can help to achieve such force feedback since it is lightweight and requires relatively little power.

¹<http://www.pokemongo.com>

²<http://anatomy4d.daqri.com>

³<https://www.aurasma.com>

⁴<https://www.microsoft.com/microsoft-hololens>

⁵<https://www.epson.de/products/see-through-mobile-viewer/moverio-bt-200>

The presented system is designed to actuate multiple muscles in parallel. For generating complex movements it is able to produce 20 different channels via time multiplexing. The system can be connected to different grid arrangements and the EMS signal parameters may be chosen freely, with some minor constraints as described in the implementation section.

Based on our hardware design and existing related work we imagine the following application scenarios.

Emotions and culture-specific behaviors: EMS can be used to transfer emotions by controlling the user's posture and gestures [9]. This can help the receiver to understand the feelings of the sender. This could also be helpful in meetings, during presentations, or in a job interview, in which a strong and convinced posture may be advantageous. Moreover, EMS in combination with a sensory system with a precise and low-latency tracking could prevent the user from performing gestures or movements that are not acceptable in a particular culture. For such scenarios multi-electrode setups are needed as many muscles have to be coordinated. For the controlled movements to look natural, fine-grained control of parameters is necessary. Moreover, the system has to be small and unobtrusive.

Learning and working: EMS can provide affordances to unknown objects [18]. Trainees could learn how to operate tools and instruments [32]. The EMS system and a tracking system could supervise and correct the movements instantaneously. The control could be transferred to the users by and by as they get better.

Augmented reality: In augmented reality high fidelity EMS feedback could extend virtual 3D objects with physical properties [26, 27]. A virtual button in front of the user could be enriched with resistance when pushing. Virtual objects like a ball could be awarded a physical weight. Other physical effects like inertia could also be simulated [7, 17].

Such scenarios need closed control loops with low reaction times. As the technical evaluation shows, the prototype system has sufficiently low reaction times. Our system is not restricted to the forearm. Kicking a virtual object with the foot [17] or pushing it with whole body could be also implemented, but would require another grid arrangement.

Smartwatches: The human body has around 700 muscles which could be used as an output medium for smartwatch notifications as in [22]. For example, the index finger may start ticking when an alarm is ringing or the foot may start shaking slightly before it starts to rain. In such scenarios different notifications may appear simultaneously.

Textile computing and on-skin technologies: To avoid the need for separate electrode grids, electrodes could be integrated in textiles or on-skin technologies [6]. It would be possible to extend sports wear, devices for rehabilitation, or clothes with electrodes to enable the presented scenarios [24]. As different clothes may have different applications and different numbers and arrangements of electrodes, connectable EMS-systems should offer flexible interfaces to interoperate with different clothes.

RELATED WORK

As our goal was to design a mobile EMS device with a large number of channels and free signal configuration options, we started with a literature survey. We found several circuit designs of stimulators in the area of medical engineering, especially in functional electrical stimulation (FES), whose goal is to restore muscle functions in paralyzed patients. The approaches and their suitability for mobile HCI scenarios differ widely. There are also different approaches to electrode grid designs and grid calibration. Related work from HCI also uses various different EMS implementations.

We first discuss different existing stimulator designs from the area of FES and take a look if they could be used as basis for an implementation of a mobile and flexible stimulator for HCI purposes. Then we present existing calibration strategies for multi-electrode EMS, because this is a requirement to use such a system. At the end we take a look at existing multi-electrode systems in HCI.

Hardware Designs of existent Stimulators

Next we present four different designs of FES stimulators and their ability to be usable for multi-electrode stimulation. Ilic et al. [11] present a microcontroller-based stimulator with four independent channels. It is designed to be a general solution for different applications in FES. It is programmable and can generate different signal patterns. The channels are galvanically isolated from the control circuit and from each other and provide biphasic stimulation currents. There are two limitations. First, the intensity of a generated pulse is not controllable via the microcontroller. There are two potentiometers for each channel, which are controlled manually. Second, the number of components is huge. A flyback DC-to-DC converter boosts the input voltage to 150 V using a transformer with eight secondary windings. Each secondary winding is a galvanically isolated voltage source of 150 V. A pair of these forms the voltage source for one channel. In this way, biphasic pulses with up to ± 150 V can be created. An extension to more channels would result in a linear growth of the number of components, which would have a negative impact on mobility.

McLeod et al. [21] designed a stimulator for the direct stimulation of motor nerves, which operates at lower voltages than on-skin surface stimulation. Their stimulator works with two 555 timer ICs to control pulse width and frequency. With an additional counter this stimulator generates pulse bursts of 1, 2, 4, or 8 pulses. The pulses get boosted up to ± 30 V with an op-amp. The intensity is controllable via a potentiometer. The voltage generation itself is not described in [21]. A limitation is that for higher voltage ranges than ± 40 V op-amps are not easily available. For this reason a simple op-amp circuit is not sufficient for on-skin stimulation, which requires higher voltages.

A circuitry for a two-channel stimulator is described in [30]. The illustrated stimulator also uses 555 timer ICs, but only generates monophasic pulses (positive pulses only), which makes it unsuitable for longer usage times. Monophasic stimulation can lead to skin irritations, because the net current is not zero. We strongly suggest using biphasic stimuli, which

do not pass net current. Moreover, with the approach described in [30] additional components are needed for every additional channel.

Therefore the before mentioned stimulators are not feasible for mobile multi-electrode systems in HCI. Malešević et al. [20] present a stimulator that creates a high voltage of about 94 V with a step-up DC-to-DC converter and creates biphasic signals with an H-bridge configuration that is not described in detail. Because the concept of stimulus generation is easy to implement, flexible and powerful, we decided to base our system on this approach. We extended the channel count to 20 without the need to modify the general hardware design. Unfortunately, Malešević et al. do not describe their circuit in detail. The H-bridge configuration is only briefly mentioned. We present the design of our own circuit in detail. Another relevant part of our design is the ability to connect different electrode grids to our stimulator. We found also here related work about existing multi-electrode systems and their calibration.

Calibration of Multi-Electrode Systems

Multi-electrode systems have been used to restore functions of the hand of patients with a palsy. One example is the restoration of grasping [20]. Popović et al. [28] investigated grid systems with selectively activatable electrodes. They found that the actuation area for controlling a specific set of muscles – which is formed by selectively activated multipad electrodes – differs from user to user, but is constant over different sessions for a single user. This means that one-time calibration of multipad electrodes is feasible per user, provided the grid is aligned in the same way as in the calibration session. In [28] the calibration is done through brute-force testing of all combinations and the measurement of resulting finger angles. The best configurations were then selected for the final control of grasp. An EMG based calibration approach has been investigated by De Marchis et al. [3] to calibrate an electrode array on the forearm to enable hand opening via EMS. They manually placed EMG Electrodes over specific muscles and measured the response of EMS stimulations that were applied by an electrode grid. The system could auto calibrate itself and could choose the best electrode configurations for hand opening.

Multi-Electrode Systems in HCI

In HCI there have been investigations of multi-electrode systems as well. *PossessedHand* [32] has shown that EMS can achieve the precise actuation of individual fingers of the hand. In contrast to [32] we present a mobile system that is more flexible and powerful. Our system can serve as a basis for replication as we offer technical details and requirements. We justify our requirements and our design decisions by taking physiological facts into account. Our system is able to produce biphasic instead of monophasic pulses [32] and offers the opportunity to adjust parameters on the fly. Also we support more electrodes than *PossessedHand* and different electrode grids can be freely connected.

Different calibration approaches have been applied in HCI. *UnlimitedHand* [31] uses photo reflectors to get information

about muscle contractions and achieves a calibration within 10 s. Electromyography (EMG) is a promising technology for the calibration of multi-electrode systems [3], because EMG delivers information about muscle contractions. Therefore, muscle positions and movements can be detected. Further EMG and EMS can share the same infrastructure [6], both technologies use electrodes that are placed at the skin area under which the targeted muscles are located. The above-mentioned FES approaches instead of [3] used external sensors like goniometers and flex sensors for calibration. These sensors were fixed to the fingers and other limbs and thus are not suitable for mobile use.

Need for Flexible Parameter Settings in HCI

In many HCI applications there is a need for multi-channel EMS systems. Lopes et al. [18] added new behaviors to physical objects through the electrical stimulation of the arm's muscles. To trigger particular object behaviors several different muscles had to be stimulated simultaneously. Moreover, there is a need for mobile stimulators that offer the ability to freely configure signal parameters like pulse width during the stimulation. Such systems would enable mobile usage of presented scenarios such as the *muscle-plotter*: "While our current version is merely portable, a wearable signal generator [...] could make muscle-plotter mobile." [19]. The used HASOMED stimulator⁶ for *muscle-plotter* is very powerful. It has 8 freely configurable channels whose stimulation parameters can be controlled in realtime. But as mentioned by the authors, the system is only "merely portable" [19]. The mobile *Let Your Body Move* toolkit [22] is able to reduce the signal intensity during the stimulation, but it cannot modify other parameters such as pulse width or frequency, because it uses commercially available stimulators as signal source. In summary, there is a need for mobile freely configurable stimulators in HCI.

REQUIREMENTS FOR MULTI-ELECTRODE EMS

From prior work by ourselves and others we identified several requirements for multi-electrode wearable EMS systems. These requirements concern anatomical factors, mobile usage aspects in terms of form factor and power consumption, the design and configurability of the multi-electrode grid, the number of parallel channels, the specification of the stimulus parameters, calibration aspects, and safety considerations.

Figure 1 shows muscles and tendons of the left upper forearm. The most muscle fibers run in the longitudinal direction of the forearm. To activate a muscle the EMS current has to travel across the muscle in the longitudinal direction. The placement of the electrode grid has to respect these anatomical factors. Figure 1 shows a possible placement and grid design consisting of two separate sleeves. Similar designs have been used in [10] and [32]. The figure also shows three active channels (3+4→12, 7→15, and 8→16) that activate three different muscles. The first channel combines electrodes 3 and 4 to form a larger pad area.

⁶<https://www.hasomed.de/en/rehamove/rehamove/product-overview.html>

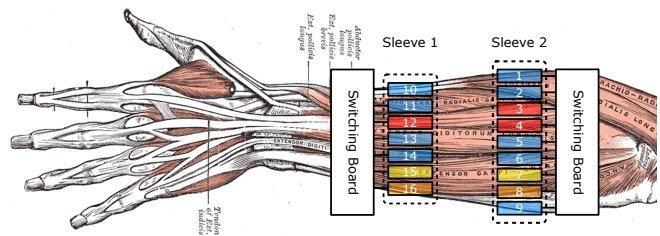


Figure 1. Anatomic figure of the left forearm. Two electrode sleeves are placed on the skin. The electrodes cover several muscles. There are three active EMS channels: 3+4→12, 7→15, and 8→16. The blue electrodes are deactivated.

One of the main advantages of EMS-based haptic feedback in contrast to mechanical feedback is the form factor. In general EMS technology is lightweight and small since it activates the motor nerves and the mechanical force is generated by the muscle fibers themselves [16]. The energy required to activate the muscles via EMS is low compared to mechanical systems, even for skin surface stimulation, which operates at a higher signal amplitude than direct nerve stimulation.

Mobile EMS systems have to be small and energy efficient. In addition to the actuation module the apparatus includes the electrodes and the switching logic that routes the EMS signal to the electrodes (see Figure 1). The wearable EMS device has to be able to communicate with other devices like mobile phones and smartwatches. The electrode grid, like other on-skin technologies [6], has to be flexible to follow the limb's contours, thin to be unobtrusive, and breathable to be comfortable for longer use. For user tests the grid should be easy to put on and sanitize. In mobile usage scenarios the EMS system should not hinder or disturb the user who is likely performing other tasks at the same time.

Human muscles are not activated sequentially to perform movements. Usually several muscles work in parallel. Natural movements result from a superposition of supporting and inhibiting muscle contractions. For example, the act of pointing the index finger somewhere involves bending the middle, ring, and pinky fingers as well as extending the index finger. In addition, several muscles are activated that control the orientations of the wrist and arm. For finger pointing, at least four muscles are activated. To create the same movement via EMS all of these muscles have to be stimulated appropriately. Complex movements were created by EMS in [9, 18]. To achieve complex movements, each muscle needs an independent electrode pair, that is placed on the skin above the muscle (see Figure 1). If the activation of the electrodes is not synchronized and multiple electrode pairs are simultaneously activated, unexpected cross currents can occur. To avoid cross currents and to achieve an independent stimulation of each muscle, the system needs as many independent channels as there are active electrode pairs.

Off-the-shelf electrodes have standard sizes and need to be customized for the shape of specific muscles and muscle groups. Smaller electrodes reduce the unwanted stimulation of neighboring muscles. However, to cover the same area with smaller electrodes means that more electrodes and more

control components are necessary, which increases the calibration effort. Effectively using a large number of electrodes calls for a calibration method that identifies the best electrode groups for the stimulation of a specific muscle and the maximum signal strength for each of these electrode groups. With a large number of electrodes, naively testing every electrode pair is not feasible. Therefore, multi-electrode systems need automatic and fast calibration algorithms. Machine learning approaches are particularly promising for calibration and for learning predefined movement patterns despite differences in anatomy and placement.

Steering limbs towards a specific posture – and staying in that posture – requires a closed-loop system [12]. The closed loop adjusts the activation of the muscle depending on observed movements. There are several parameters in addition to signal on- and off-time that are helpful in closed-loop control. These include the selection of the active electrodes, the current intensity, the pulse frequency, the pulse width, and the inter-phase pause. No commercial mobile device that we know of allows controlling all of these stimulation parameters. In most mobile, commercially available stimulators pulse frequency, pulse width, and intensity can be adjusted once before the stimulation starts, but adjustment during the stimulation is not normally possible. In contrast not mobile systems like the HASOMED *RehaMove* medical stimulator enable parameter control during stimulation by its integrated ScienceMode2⁷ protocol. Furthermore, a freely configurable system may be used to reduce muscle fatigue by changing the stimulation parameters or by time-division multiplexing the activated electrodes [4]. A freely configurable system also allows choosing parameters so as to optimize user experience and comfort.

In closed-loop systems the response time of the EMS devices needs to be low to adjust the movements of the limbs or to generate a counter force to the user's movements. The response time has different components: the delay of the EMS device, the time that the muscles need to respond (fast-twitch muscle fibers typically in 20 ms to 50 ms, slow-twitch muscle fibers 60 ms to 120 ms [15]), and the system that tracks the movement and position of the limbs to adjust the EMS signal parameters.

Multi-electrode systems have to guarantee technical safety for particular cases. For example, if stimulation is applied to both arms the system has to ensure that both stimulations are mutually exclusive. A current flow through the chest must never occur. Moreover, the maximum stimulation current must not be exceeded. The maximum comfortable current depends on the electrode size and the sensitiveness of the user. If the electrode sizes are known to the system, it can compute the current density (current per area) and prohibit the application of large currents to small areas.

CONCEPTUAL DESIGN

This section gives a high-level overview of the conceptual design of our multi-channel wearable EMS system. In the next

⁷<https://www.hasomed.de/en/rehamove/rehamove/science-mode.html>

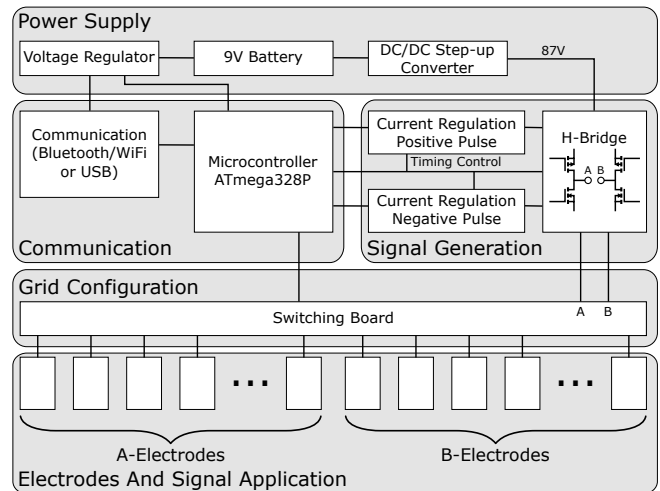


Figure 2. Simplified block diagram of the EMS system for 20 freely configurable EMS channels.

section we present the details of our implementation of this conceptual design. The main components and functions of the system include (see also Figure 2):

- **Power supply.** Current mobile devices are battery powered. EMS requires relatively high voltages at low currents. A step-up converter transforms the low voltage of the battery to a high voltage for stimulation. The overall power consumption for generating the EMS signal is low compared to vibration motors or other force feedback technologies.
- **Signal generation.** Biphasic EMS signals consist of a positive pulse followed by a negative pulse. The pulses are relatively short (typically 100 μ s) compared to the period (typically 10 ms). There are multiple independent time-multiplexed signal channels without cross currents. Amplitude, frequency, pulse widths, and inter-phase pause are configurable in real time. To ensure charge compensation for long-term usage, EMS signals should not be monophasic (only have a single polarity), because that could lead to skin irritations.
- **Grid configuration.** Different positions of application on the human body require different layouts and electrode configurations. The system has to allow different layouts and different activation patterns. There needs to be a switching logic to distribute the signal to the currently active electrode pair.
- **Electrodes and signal application.** Different positions of signal application on the human body require different electrode shapes. Depending on the applications different materials are suited best. Electrode variants include rubber carbon, self-adhesive, and conductive gel electrodes.
- **Communication.** The EMS control device controls the signal generation unit and configures the electrode grid. Standard low-power wireless technologies such as Bluetooth LE are well suited for communication between the EMS control device and commodity devices with a user interface, such as smartwatches and mobile phones.

- *Calibration.* Multi-electrode grids are particularly difficult to calibrate. Calibration involves identifying the maximum signal strength that is still comfortable for the user and finding the best grid position and electrode activation pattern to achieve the desired effect.
- *Feedback loop.* Precise control requires a feedback loop. Such a feedback loop can be implemented with external sensors and tracking devices. In this paper we do not cover sensing technologies for implementing feedback loops.

HARDWARE PROTOTYPE

Our implementation of the hardware prototype is based on a ATmega328P⁸ microcontroller, which communicates to PCs via USB or to mobiles over Bluetooth (Figure 2). The microchip was programmed with the Arduino Eclipse IDE⁹. The prototype is powered by a 9 V battery for mobile usage and has a step-up converter, which produces high voltage for the current-controlled muscle stimulation. The biphasic pulses to stimulate the muscles are generated by an H-bridge, which is controlled by the microcontroller and current regulation circuits as depicted in Figure 2.

The prototype has one physical channel, which is time multiplexed by a switching board (Figure 2). The switching board and the signal generation circuit operate at 2 kHz to generate 20 channels at 100 Hz with a pulse width of 50 μ s. Intensity and timing of the stimulation signal are controlled by the H-bridge. The switching board controls the application area of the signal by activating and deactivating electrodes. This happens after each generated biphasic pulse. In this way the system is able to control 20 channels in parallel and can stimulate each of the 20 active electrode pairs with a different stimulation intensity. This time-division multiplexing of a single physical channel prevents unintended cross currents between electrode pairs that happen to be activated at the same time.

EMS parameters like pulse width, inter-phase pause, pulse frequency, and pulse intensity are all adjustable in real time according to our requirements. In the current prototype, the maximum output amplitude of a pulse is limited to 104 mA. Each pulse is generated independently and individually. Time multiplexing and electrode discharging ensures that every virtual channel is independent of every other virtual channel.

The grid is designed to control hand gestures (compare Figure 9). The muscles that effect hand gestures are located in the lower arm. The direction of the muscle fibers of most of these muscles corresponds to the longitudinal direction of the lower arm (see Figure 1). To be able to actuate these muscles we split the electrode grid into two sleeves with 22 and 18 electrodes, respectively. The lower sleeve has fewer electrodes, because the lower forearm has a smaller diameter than the upper forearm. The electrodes are molded in silicone. It is also possible to design other grid layouts, which cover the full forearm or are designed for other body locations.

⁸<http://www.microchip.com/wwwproducts/en/ATmega328P>

⁹<http://www.baeyens.it/eclipse/>

Our two-sleeve layout is similar to the ones in *NESS Handmaster* [10] and *PossessedHand* [32]. In contrast to these systems our system can run 20 channels in parallel and each pulse can be controlled individually. We use more electrodes with a smaller width to enable a more fine-grained control. Our system can combine an arbitrary number of electrodes to form a larger area. For example, in Figure 1 electrodes 3 and 4 are combined. The properties of the silicone ensure that the sleeves are sufficiently flexible and elastic to follow the shape of the forearm. We use commercially available carbon rubber electrodes¹⁰ that are easy to sanitize and are reusable for many times. The electrodes were cut to a size of 1×4 cm with a laser cutter, placed into a 3D-printed mold, and covered with liquid silicone.

As a power supply for the signal generation circuit we used a MAX773¹¹ step-up converter, which boosts the 8.4 V of the battery to 87 V. Capacitors are used for short-term energy storage. The circuit of the step-up converter is based on the circuit in Figure 3e of the MAX773 data sheet. There are a few differences compared to our circuit: We do not use a shunt resistor, because the input voltage is in the recommended region for powering the MAX773. Furthermore, the resistor values differ, because we do not boost the voltage to 100 V but only to 87 V. Recommended resistor values for different voltages are described in the data sheet.

As shown in Figure 3 the H-bridge consists of two P-MOSFETs (1 & 2) and two N-MOSFETs (3 & 4) for the signal generation. The P-MOSFETs are connected to the 87 V output of the step-up converter, the N-MOSFETs are connected to ground. The connection to the EMS electrodes forms the cross-bar. To create a pulse, two diagonal MOSFETs have to be activated. The red path over MOSFETs 1 and 4 generates positive pulses, the blue path over MOSFETs 2 and 3 generates negative pulses (Figure 3). The pulse length is controlled by the on-time of the corresponding MOSFETs in the path. The EMS stimulus intensity is controlled by the current strength of the path. There are also additional grounding MOSFETs parallel to the N-MOSFETs, which discharge the electrodes after a stimulation pulse.

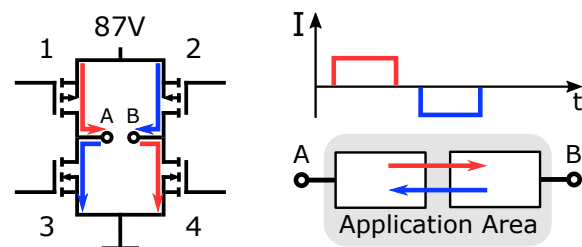


Figure 3. H-bridge: Controls the positive and negative edges of the EMS signal.

The intensity regulation is coupled with the H-bridge (cf. Figure 2). The regulation circuit is shown in Figure 4. For each

¹⁰<http://www.medtronic.com/covidien/products/pain-therapy-recovery/uni-patch-specialty-electrodes>

¹¹<https://datasheets.maximintegrated.com/en/ds/MAX770-MAX773.pdf>

individual pulse the current can be regulated linearly in 256 steps and in a range of 0 to 104 mA. The current is controlled by an op-amp, which forms a constant current source with a N-MOSFET of the H-bridge, a small-value measurement resistor R_m and a digital potentiometer. The potentiometer is in series with another resistor and forms a voltage divider. The voltage at the wiper of the potentiometer in Figure 4 is the reference voltage for the current regulation. The op-amp operates with a negative feedback circuit. It compares the voltage across the measurement resistor with the given reference voltage and increases or decreases the resistance of the N-MOSFET so as to get the same voltage across the measurement resistor. This setup works as a constant current source as long as the voltage of 87 V is sufficient to generate the desired current. The current through the measurement resistor, which also flows through the electrodes, can be calculated with Ohm's law as $I_{EMS} = \frac{U}{R}$. Knowing the current is useful when reporting user studies or comparing generated forces and effects with other experiments. In our system the applied current is always known.

The generation of a biphasic stimulation pulse has five phases: First both digital potentiometers of both current regulation circuits get new values. Then in the second phase the positive pulse is generated with the activation of MOSFETs 1 and 4 (Figure 3). MOSFET 4 is activated when voltage is applied to the voltage divider in Figure 4 that consists of a resistance and a digital potentiometer. The current begins to flow on the red path (Figure 3). The op-amp regulates the current to the desired value. When the pulse on-time has passed both MOSFETs are deactivated and the current stops. Now the third phase, the inter-phase pause, begins. The grounding MOSFET on clamp *A* is activated to discharge any electrodes that are connected to *A*. After the inter-phase pause the grounding MOSFET is deactivated. The fourth phase is the generation of the negative pulse. For this phase MOSFETs 2 and 3 are activated and current regulation happens via MOSFET 3. The stimulation current flows over the blue path in Figure 3. After the on-time of the negative pulse MOSFETs 2 and 3 are deactivated. In the fifth and final phase the electrodes that are connected to clamp *B* are discharged by the grounding MOSFET, which is connected to clamp *B*. After grounding the electrodes the generation of one biphasic pulse is complete. If so configured the grid configuration now changes to connect another electrode pair with the signal generation unit or it disconnects all electrodes. After that the signal generation unit can create the next stimulus with a new parameter set.

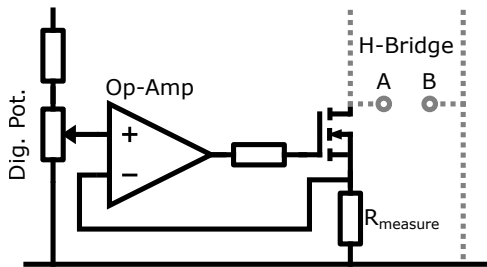


Figure 4. Current regulator: Regulates the amplitude of the EMS signal.

The crossbar of the H-bridge is connected to the switching board via clamps *A* and *B*. It directs the stimulus current to the desired electrodes (Figure 2). In our approach the switching board consist of three serially connected identical switching boards. One of our switching boards consists of a 16-channel LED driver and 16 photocouplers. Each photocoupler is connected to the LED driver and can be switched on or off by the LED driver. If a photocoupler is activated the corresponding electrode gets connected to the H-bridge. An electrode could be connected either to clamp *A* or *B* (Figure 3). This way electrodes are not bonded to specific virtual channels. Each electrode could be used with each channel. Current only flows if there is at least one electrode connected to *A* and one electrode connected to *B*, as shown in Figure 3. For maximum flexibility electrodes should be connectable to both clamps, which would require two photocouplers per electrode. In our implementation we did not include a second photocoupler for every electrode as this would have significantly increased the complexity and doubled the number of components. As a trade-off between flexibility and number of components we integrated a jumper for every group of 8 electrodes. Via the jumper a group of eight photocouplers can be connected to either *A* or *B*. In our two sleeve prototype the switching boards are configured the way that electrodes of the lower sleeve are only connectable to *A* and the electrodes of the other sleeve are only connectable to *B*. Connecting multiple electrodes at the same time as shown with electrodes 3 and 4 in Figure 1 is also possible. This way large electrode areas can be created by combined activation of two or more electrodes. The switching board can change the electrode configuration with each cycle. The maximum switching rate is $\approx 200 \mu s$ (fastest switching rate of the photocoupler TLP222G¹²). The communication with the switching boards is done serially, because each LED driver has an internal 16-bit shift register.

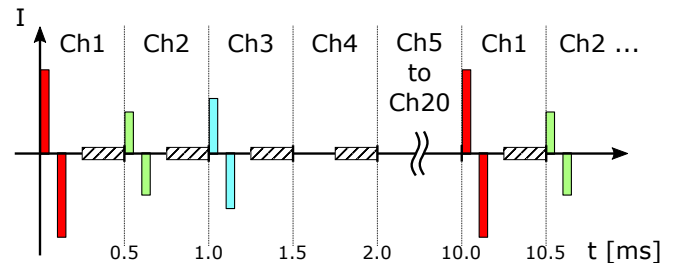


Figure 5. Time multiplexing of 20 channels, each running at 100 Hz with 50 μs pulse width. Channels 1, 2, and 3 use different currents. The other channels are disabled. The hatched areas denote times that are used for grid reconfiguration.

We propose a constant frequency for all virtual channels (time slot length is equal for every virtual channels), to make the system easy to program. In our example approach we set the frequency to 100 Hz. This common frequency for all virtual channel is freely adjustable. There are a number of additional considerations regarding the time multiplexing as the system provides only one physical channel:

¹²<https://toshiba.semicon-storage.com/info/docget.jsp?did=17038&prodName=TLP222G-2>

First the maximum channel count of the stimulator depends on the frequency and the time slot length for each channel: $\text{channelCount} = \frac{1}{f_{\text{EMS}} \times \text{timeSlotLength}}$. In our example approach each channel runs at frequency of 100 Hz and each time slot has a duration of 500 μs (compare Figure 5). This ends up in a amount of 20 channel: $\text{channelCount} = \frac{1}{100 \text{ Hz} \times 500 \mu\text{s}} = 20$. Lowering the frequency enables either more virtual channels (time slot length is kept equal) or longer time slots (channel count is kept equal). By increasing the overall stimulation frequency either the amount of virtual channels decreases (time slot length is kept equal) or the time slot length decreases (channel count is kept equal). There are no “predefined” frequencies for the stimulator, but different choices of frequencies entail different dependencies and restrictions to the amount of channels and the time slot length, that restricts the pulse width of the positive and negative pulse as the pause between them. E.g. decreasing the frequency to 50 Hz for all virtual channels and keeping the time slots length equal doubles the possible channel count to 40.

Second each virtual channel has a time slot, that is scheduled in a round robin fashion (as depicted in Figure 5). The time slot length has to be longer than the sum of the times for the negative pulse t_{neg} , the positive pulse t_{pos} , the inter-phase pause t_{pause} and the grid reconfiguration time t_{grid} : $\text{timeSlotLength}_{\text{min}} = t_{\text{pos}} + t_{\text{neg}} + t_{\text{pause}} + t_{\text{grid}}$. If the stimulation signal is longer than the given time slot the frequency of the overall system decreases. This may happen with increasing the pulse duration and/or the inter-phase pause beyond certain values. Limited by our used hardware, 250 μs of each time slot are needed for the reconfiguration of the electrode grid (200 μs for switching the photocouplers and 50 μs buffer). Therefore it does not make sense to decrease the time slot length to 250 μs or smaller values. We set our parameter to the following values: $t_{\text{pos}} = t_{\text{neg}} = t_{\text{pause}} = 50 \mu\text{s}$ and 100 μs of the 500 μs are not used (compare Figure 5).

This interdependence of the parameters is a restriction but not a severe one. First an optimal stimulation frequency could be set for all channels. Second there are different possibilities to change the frequency of virtual channels running them at higher or lower frequencies than the first predefined:

If necessary, a lower frequency for each virtual channel is possible by dropping biphasic pulses for this channel. After the creation of a pulse the next n pulses of the channel could be dropped at runtime. The resulting frequency would be: $\frac{f_{\text{EMS}}}{(n+1)}$. For an exemplary stimulation frequency of 100 Hz frequencies of 50 Hz, 33 Hz, 25 Hz and so on are possible. Moreover, the frequency of the complete system could be changed to desired values (e.g., every EMS channel could run at 80 Hz).

Higher frequencies like 200 Hz, as used in *muscle-plotter* [19], could be achieved by combining 2 channels: Channels 1 and 11 could be combined to actuate the same application area with the same parameters. Also much longer pulse widths could be handled with our system by combin-

ing 2 or more sequential channels. By combining 2 sequential channels 750 μs are available for the biphasic signal. The remaining 250 μs are reserved for grid reconfiguration. Yet, in such configurations the number of parallel channels decreases.

If the system is used for a single channel, all parameters can be adjusted freely. This is in principle also possible with multiple parallel channels, but arbitrary parallel channels with different frequencies lead to non-trivial scheduling problems. In this case there are different tasks (channels) with different durations (channel widths/time slot durations) that have different frequencies and have hard real-time requirements on one physical CPU (signal generation unit). To avoid the need for implementing complex scheduling algorithms we therefore recommend using the system with predefined frequencies and time slots.

Before we continue to discuss the restriction of an overall constant frequency, we should turn to physiological facts related to EMS frequency. The minimum EMS frequency is at about 30 Hz. Lower frequencies will not guarantee bringing the muscle to a tetanus (sustained contraction over time), but lead to twitching in the frequency of the generated signal. Higher forces can be created with higher frequencies [5], but the muscles fatigue more quickly with higher frequencies. As a tradeoff of channel count, frequency, channel width, force, and muscle fatigue we decided to use 20 channels at 100 Hz. However, the system is not restricted to these values. E.g. 80 Hz or 120 Hz are possible, pulse widths of 300 μs or 10 μs are also possible. But the amount of channels depends on all settings as mentioned above.

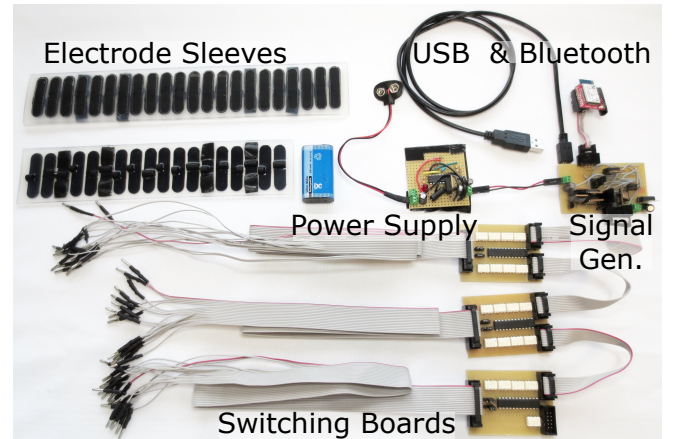


Figure 6. Implementation of our hardware prototype. Power supply and signal generation are physically separated to two PCBs. The switching boards could be serially extended as needed for the desired application. Different electrodes could be connected to the switching boards with 2 mm pin headers.

TECHNICAL EVALUATION

The evaluation is structured according to the requirements formulated above: anatomical factors, mobile usage aspects in terms of form factor and power consumption, the design and configurability of the multi-electrode grid, the number of parallel channels, the specification of the stimulus parameters, calibration aspects, and safety considerations.

Our current prototype fulfills the above-mentioned requirements regarding the form factor and power consumption. Figure 6 shows the current implementation of our prototype. It can be carried in a small pocket or placed on the lower arm. The system consists of three switching boards of $6.5 \times 4.7 \times 2.0$ cm (width \times depth \times height), a control and communication board of $7.3 \times 4.2 \times 3.0$ cm, and a power board of $5.5 \times 5.0 \times 2.5$ cm. The sleeves are 28×5 cm 23×5 cm, respectively, and have a height of about 0.9 cm.

The prototype's battery lasts for several hours. In standby mode, with Bluetooth switched on but not connected, it consumes about 90 mA. If Bluetooth is on and connected, the prototype consumes about 150 mA typically and up to 320 mA if all channels are on and set to maximum current. Usually the actuation times are quite short and not all muscles are activated in parallel. As EMS leads to muscle fatigue a continuous stimulation is not feasible anyway. In a typical scenario, in which half of the channels are on at 50% of the maximum intensity an 8.4 V battery with a capacity of 270 mAh achieves an estimated usage time between 70 minutes and 3 hours. There are compact batteries with much higher capacity. Lithium-ion batteries like the ones used in smartphones have five times of the capacity of our used battery (8.4 V battery: 2.27 Wh; modern smartphone battery: 10.0 Wh and more). In addition, the replacement of some of the components with a focus on power consumption would likely increase the mobile usage time of the prototype even further: the current prototype uses standard Bluetooth (not Bluetooth LE) and the power supply uses linear voltage regulators to provide the necessary input voltages.

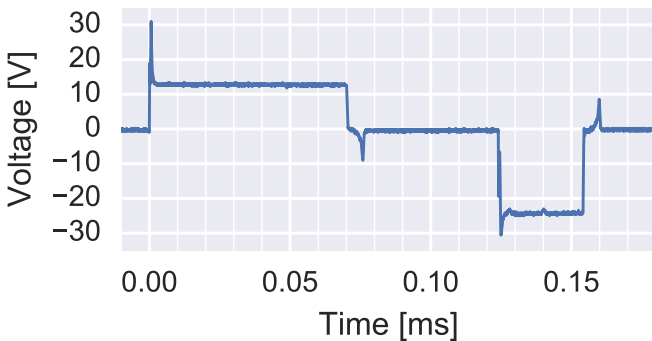


Figure 7. Oscilloscope probe of generated pulse over 220Ω . The pulses are not perfectly rectangular. The durations are: $70 \mu\text{s}$ for the positive pulse, $50 \mu\text{s}$ for the inter-phase pause, and $30 \mu\text{s}$ for the negative pulse. The positive pulse has half the maximum intensity of $12 \text{ V} / 220 \Omega = 55 \text{ mA}$. The negative pulse has a maximum intensity of 110 mA .

Figure 7 shows an oscilloscope probe of freely configured biphasic pulses generated by the system. The intensity, on-time, and inter-phase pause may be adjusted for each pulse. Moreover, different current intensities may be chosen for each channel, as shown in Figure 8. The figure shows different intensities for five different channels. The intensity is adjustable in 256 steps from 0 mA to 104 mA. The length of each time slot is $500 \mu\text{s}$ (Figure 5). At a stimulation frequency of 100 Hz 20 channels fit in the period of 10 ms.

As Figure 7 shows there are very brief spikes at the onset of a pulse and when discharging the electrodes. The spikes are not perceptible and do not create a muscle contraction or sensation, because of their short duration of $\approx 1 \mu\text{s}$.

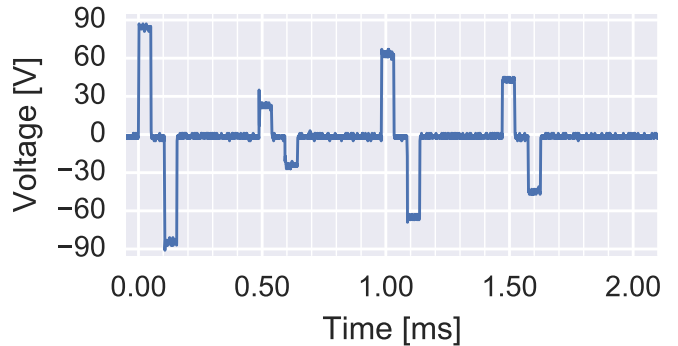


Figure 8. Oscilloscope probe of 5 virtual channels with pulse widths of $50 \mu\text{s}$ measured over a resistive load of 750Ω . The channels have the intensities 100%, 25%, 75%, 50%, and 0%, respectively. The length of the time slots is $500 \mu\text{s}$.

To evaluate the performance of our system we measured the reaction time for two common scenarios: The first scenario is the activation of an unused channel. This scenario consisted of setting the grid configuration, setting the intensity, and activating the desired channel. The second scenario consists of changing the intensity or grid configuration of an active channel. The first scenario represents the reaction of the system to a new stimulation and the second scenario represents the time of parameter changes during the stimulation. Both times should be small for a closed loop control. We measured the time it took to send the commands, apply the settings, generate the first biphasic pulse with the applied settings, send an acknowledgement, and receive it.

In the first scenario three commands have to be sent to the device. The measured mean time was 45.30 ms with a standard deviation of $\pm 7.09 \text{ ms}$. The measured mean time of sending one command for the second scenario was 28.27 ms with a standard deviation of $\pm 6.98 \text{ ms}$. As these mean times include the time needed for sending back the acknowledgment, the real reaction time of the system is shorter. In a subsequent test we measured the time of sending one byte to the device and sending an acknowledgment byte back. The mean time was 21.43 ms with a standard deviation of $\pm 6.59 \text{ ms}$. Therefore we assume that the acknowledgment takes about 10 ms to be sent and transferred back to our measurement application and the overall mean reaction time of the system is about 35 ms.

One important aspect regarding safety is that our prototype is battery powered. There is no common ground with items of the surrounding environment. Also the maximum current draw is limited by the battery through its capacity and resistance. The stepped up voltage of 87 V is provided by capacitors. If the battery is disconnected the capacitors discharge. The H-bridge components are rated for much higher voltages (up to 200 V) and currents (up to 3.3 A). Therefore they are safe as the voltage and currents are in ranges up to 87 V and 104 mA. Unwanted cross currents between different channels

are avoided, because there is only one physical channel that produces only one signal at any specific point in time. In addition, there are no unwanted crosscurrents between different electrodes, because only the desired electrodes are connected to the stimulation generator and all others are disconnected. To ensure safety over the complete usage time, we enabled the internal brown-out detection and the watchdog of the microcontroller. The internal brown-out detection detects voltage drops, that may be triggered by a short circuit or an empty battery. The watchdog timer resets the microcontroller if it gets not resetted by the running program. This may happen if the program crashes. In the case of a voltage drop below 4,5 V or in the case of a missing watchdog reset for more than 15 ms the microcontroller resets and the circuitry transfers into a safe state. Also if the Bluetooth connection is lost, our system will deactivate all channels after one second and will transfer into a safe state. As with any EMS device, all safety instructions for such systems have to be adhered to. [30] provides a list of general safety instructions for EMS devices. Note that at the moment our system is not medically approved. The certification is in progress. Anyway we tested

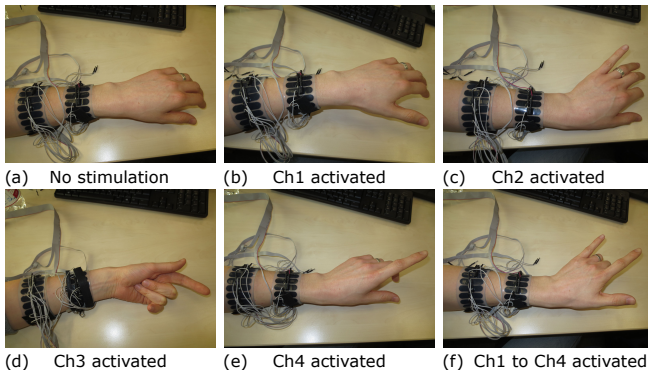


Figure 9. The two sleeves worn at the forearm. The upper sleeve has 22 electrodes, the lower sleeve has 18 electrodes. Part (a) shows the hand without stimulation. Parts (b) to (e) each show the hand when one channel is activated. Each channel activates different electrodes that stimulate different muscles. Channel 1 actuates the extensor of the thumb. Channel 2 extends the pinky finger. Ring and middle finger are bent by an activation of channel 3. Channel 4 extends the index finger. Part (f) shows the combined movement if all channels are on simultaneously.

the system on ourselves. Figure 9 shows an example application of our prototype. The parallel stimulation of different muscles allows us to create complex hand gestures like (f). Figure 9 (b) to (e) each shows the hand movement regarding a single stimulation that is triggered by one channel (channel 1 to 4). If all four channels are active at the same time, the hand makes a merged posture (f) that is a combination of the four single movements. For a comparison of the effect of each channel Figure 9 (a) shows the hand, when no stimulation is applied.

CALIBRATION

Calibration for grid systems is challenging. Theoretically with our two sleeve system (Figure 9) with a set of 18 *A* electrodes and 22 *B* electrodes there are about 1.1×10^{12} possible electrode configurations that could be tested.

This number reduces to 396 configurations (the Cartesian product) if only one electrode from set *A* and one from set *B* are activated at the same time. We do not need to test merged electrodes like $(1+2) \rightarrow (10+11)$ (Figure 1), because the effect of merged electrodes can be achieved by selectively activating the single electrode pairs that make up the combination $(1 \rightarrow 10 + 2 \rightarrow 11)$. The effect of the sequentially activated single electrode pairs will superpose if the time between the activation of the individual pairs is sufficiently short [28]. If there are enough free channels, a sequential stimulation of more electrode pairs should be preferred to an activation of large merged electrodes, because this reduces muscle fatigue [20].

However, not all of these remaining possible configurations make sense. *PossessedHand*, for example, only uses 14 configurations: Each sleeve has 14 electrodes and only electrodes directly opposite of each other were tested [32].

In contrast we suggest testing each upper sleeve electrode with 5 different electrodes of the lower sleeve as shown in Figure 10. This results in $22 \cdot 5 = 110$ combinations. We think that this reduction is valid, because most of the muscles in forearm are aligned in the longitudinal direction of the forearm (cf. Figure 1). If a muscle runs slightly off the longitudinal direction, it is still covered by one of the combinations shown in Figure 10.

For each of these combinations the maximum comfortable intensity has to be determined. In the calibration process, for each of the configurations the current is increased until the maximum intensity is reached. Resulting limb movement can be measured by an optical systems or any other sensory system (flex sensors, goniometers, optical systems like optitrack, leap motion, etc.). At the end of process, best configurations for specific limb movements can be selected.

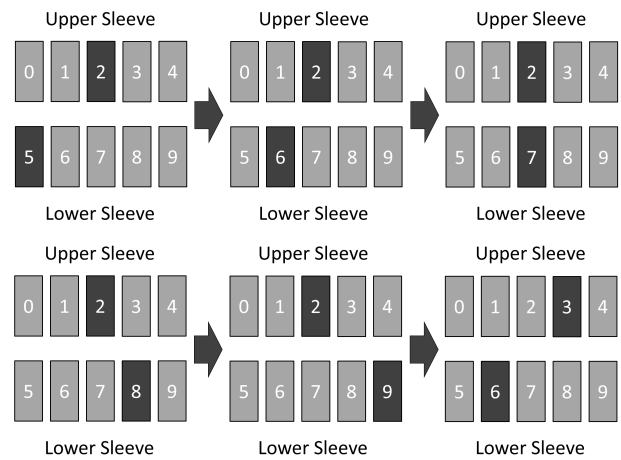


Figure 10. Calibration pattern for two-sleeve system. For calibration, electrode 2 is combined in turn with electrodes 5 to 9, forming the channels $2 \rightarrow 5$, $2 \rightarrow 6$, $2 \rightarrow 7$, $2 \rightarrow 8$, and $2 \rightarrow 9$. Then calibration proceeds with electrode 3 of the upper sleeve.

The calibration process is time consuming, irrespective of the used sensory system. Strategies to overcome this limitation need to be found to enable scenarios in a mobile and everyday usage scenario. Popović and Popović claim that electrode

configuration does not change for a single user over different sessions [28]. By placing the electrode grid at the same position only the stimulation intensities have to be recalibrated. This is possible, but it cannot be guaranteed that the grid is aligned in the same orientation as in the calibration session. Systems need sensory input to detect their placement. *UnlimitedHand* uses photo-reflectors for the calibration of the system. After the calibration the control and recognition of four hand gestures is possible.

As our system is not medically tested we only tested the system on ourselves and found that the maximum comfortable intensity is relatively stable over multiple sessions. We also found a session-independent intensity profile on the lower arm that could be used to enhance the calibration process. But these initial findings have to be validated in a full user study once the system is medically approved.

CONCLUSIONS

In mobile HCI there is a need for wearable haptic feedback technologies. EMS could be such a technology, in particular when designed with flexible, multi-electrode grids. Complex movements created by investigations like *Emotion Actuator* [9] or *Affordance++* [18] would clearly benefit from such general and powerful EMS systems. However, even seemingly simple natural movements often require the coordinated actuation of multiple muscles.

Common mobile stimulators limit research because of fixed parameter settings during stimulation and a limited channel count. The parallel stimulation of more muscles is only possible by adding further stimulators, which becomes bulky at some point. Therefore there is a need for mobile multi-channel devices like the system we propose. General systems are needed that can be connected to different grid layouts for different body positions (e.g. the lower extremities or the upper arm). Researchers need open platforms to have full control over parameters and application areas. EMS actuation and sensory systems like *UnlimitedHand* [31] are milestones on the way to general EMS systems.

For these reasons we investigated an enabling technology platform for non-trivial EMS actuation applications. We presented a prototype of an EMS stimulator that is designed according to requirements of mobile HCI. This includes the stimulation current generator, the switching board, and the design of the electrode grid. The working prototype was successfully evaluated technically against the formulated requirements. We present the system design, the design considerations, and evaluation results in detail. The three parts of the hardware design may be used as a complete system or combined with other existing systems (stimulators as well as other electrode grids or switching boards). We provide the schematics online¹³ to enable replication of the system.

The main limitation is that the prototype is not medically certified yet. But we are in progress. Furthermore, multi-electrode systems require more advanced calibration algorithms and strategies. *UnlimitedHand* [31] shows an example of a fast calibration for 11 electrodes. Findings from medical

applications of electrode grids should be considered and used in HCI as well.

A software library for the stimulator has to be developed that provides a simple interface to the configuration of channels and their parameters. It should validate the user input and ensure that only possible configurations are actually set.

Our next steps towards the envisioned application scenarios will be the implementation of closed-loop control [12] with EMS and investigations of easy-to-use automatic calibration methods for EMS. As mentioned, we think about a combination of EMS and EMG to turn the user's muscles into an input and output interface.

REFERENCES

1. Badshah, A., Gupta, S., Morris, D., Patel, S., and Tan, D. Gyrotab: A handheld device that provides reactive torque feedback. In *Proceedings of the SIGCHI Conference on Human Factors in Computing Systems, CHI '12*, ACM (New York, NY, USA, 2012), 3153–3156.
2. Bowman, B. R., and Baker, L. L. Effects of waveform parameters on comfort during transcutaneous neuromuscular electrical stimulation. *Annals of biomedical engineering* 13, 1 (1985), 59–74.
3. De Marchis, C., Santos Monteiro, T., Simon-Martinez, C., Conforto, S., and Gharabaghi, A. Multi-contact functional electrical stimulation for hand opening: Electrophysiologically driven identification of the optimal stimulation site. *Journal of NeuroEngineering and Rehabilitation* 13, 1 (2016), 22.
4. Downey, R. J., Bellman, M. J., Kawai, H., Gregory, C. M., and Dixon, W. E. Comparing the induced muscle fatigue between asynchronous and synchronous electrical stimulation in able-bodied and spinal cord injured populations. *IEEE Transactions on Neural Systems and Rehabilitation Engineering* 23, 6 (Nov 2015), 964–972.
5. Dreibati, B., Lavet, C., Pinti, A., and Poumarat, G. Influence of electrical stimulation frequency on skeletal muscle force and fatigue. *Annals of physical and rehabilitation medicine* 53, 4 (2010), 266–277.
6. Duentz, T., Pfeiffer, M., and Rohs, M. On-skin technologies for muscle sensing and actuation. In *Proceedings of the 2016 ACM International Joint Conference on Pervasive and Ubiquitous Computing: Adjunct, UbiComp '16*, ACM (New York, NY, USA, 2016), 933–936.
7. Farbiz, F., Yu, Z. H., Manders, C., and Ahmad, W. An electrical muscle stimulation haptic feedback for mixed reality tennis game. In *ACM SIGGRAPH 2007 posters*, ACM (2007), 140.
8. Hassan, M., Daiber, F., Wiehr, F., Kosmalla, F., and Krüger, A. Footstriker: An EMS-based foot strike assistant for running. *Proc. ACM Interact. Mob. Wearable Ubiquitous Technol.* 1, 1 (Mar. 2017), 2:1–2:18.

¹³<http://hci.uni-hannover.de/research/zap++>

9. Hassib, M., Schneegass, S., Pfeiffer, M., Rohs, M., and Alt, F. Emotion Actuator: Embodied emotional feedback through electroencephalography and electrical muscle stimulation. In *Proceedings of the 35th Annual ACM Conference on Human Factors in Computing Systems*, CHI '17 (New York, NY, USA, 2017).
10. Ijzerman, M. J., Stoffers, T. S., In't Groen, F., Klatte, M. A. P., Snoek, G. J., Vorsteveld, J. H. C., Nathan, R. H., and Hermens, H. J. The NESS Handmaster Orthosis: Restoration of hand function in C5 and stroke patients by means of electrical stimulation. *Journal of rehabilitation sciences* 9, 3 (1996), 86–89.
11. Ilic, M., Vasiljevic, D., and Popovic, D. B. A programmable electronic stimulator for FES systems. *IEEE Transactions on Rehabilitation Engineering* 2, 4 (Dec 1994), 234–239.
12. Jagacinski, R. J., and Flach, J. M. *Control theory for humans: Quantitative approaches to modeling performance*. CRC Press, 2003.
13. Jones, D., Bigland-Ritchie, B., and Edwards, R. Excitation frequency and muscle fatigue: Mechanical responses during voluntary and stimulated contractions. *Experimental neurology* 64, 2 (1979), 401–413.
14. Kaul, O. B., Pfeiffer, M., and Rohs, M. Follow the force: Steering the index finger towards targets using EMS. In *Proceedings of the 2016 CHI Conference Extended Abstracts on Human Factors in Computing Systems*, CHI EA '16, ACM (New York, NY, USA, 2016), 2526–2532.
15. Knudson, D. *Fundamentals of Biomechanics*. Springer Science & Business Media, 2003.
16. Lopes, P., and Baudisch, P. Muscle-propelled force feedback: Bringing force feedback to mobile devices. In *Proceedings of the SIGCHI Conference on Human Factors in Computing Systems*, ACM (2013), 2577–2580.
17. Lopes, P., Ion, A., and Baudisch, P. Impacto: Simulating physical impact by combining tactile stimulation with electrical muscle stimulation. In *Proceedings of the 28th Annual ACM Symposium on User Interface Software & Technology*, UIST '15, ACM (New York, NY, USA, 2015), 11–19.
18. Lopes, P., Jonell, P., and Baudisch, P. Affordance++: Allowing objects to communicate dynamic use. In *Proceedings of the 33rd Annual ACM Conference on Human Factors in Computing Systems*, ACM (2015), 2515–2524.
19. Lopes, P., Yüksel, D., Guimbretière, F., and Baudisch, P. Muscle-plotter: An interactive system based on electrical muscle stimulation that produces spatial output. In *Proceedings of the 29th Annual Symposium on User Interface Software and Technology*, ACM (2016), 207–217.
20. Malešević, N. M., Maneski, L. Z. P., Ilić, V., Jorgovanović, N., Bijelić, G., Keller, T., and Popović, D. B. A multi-pad electrode based functional electrical stimulation system for restoration of grasp. *Journal of NeuroEngineering and Rehabilitation* 9, 1 (2012), 66.
21. McLeod, K., Lovely, D., and Scott, R. A biphasic pulse burst generator for afferent nerve stimulation. *Medical and Biological Engineering and Computing* 25, 1 (1987), 77–80.
22. Pfeiffer, M., Dunte, T., and Rohs, M. 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*, ACM (2016), 418–427.
23. Pfeiffer, M., Dunte, T., Schneegass, S., Alt, F., and Rohs, M. Cruise control for pedestrians: Controlling walking direction using electrical muscle stimulation. In *Proceedings of the 33rd Annual ACM Conference on Human Factors in Computing Systems*, CHI '15, ACM (New York, NY, USA, 2015), 2505–2514.
24. Pfeiffer, M., and Rohs, M. Haptic Feedback for Wearables and Textiles based on Electrical Muscle Stimulation. In *Handbook of Smart Textiles and Textile Electronics*. Springer HCI Series.
25. Pfeiffer, M., Schneegass, S., Alt, F., and Rohs, M. Let me grab this: A comparison of EMS and vibration for haptic feedback in free-hand interaction. In *Proceedings of the 5th augmented human international conference*, ACM (2014), 48.
26. Pfeiffer, M., Schneegass, S., Alt, F., and Rohs, M. Let me grab this: A comparison of EMS and vibration for haptic feedback in free-hand interaction. In *Proceedings of the 5th Augmented Human International Conference*, AH '14, ACM (New York, NY, USA, 2014), 48:1–48:8.
27. Pfeiffer, M., and Stuerzlinger, W. 3D virtual hand selection with EMS and vibration feedback. In *Proceedings of the 33rd Annual ACM Conference Extended Abstracts on Human Factors in Computing Systems*, CHI EA '15, ACM (New York, NY, USA, 2015), 1361–1366.
28. Popovi, D. B., and Popovi, M. B. Automatic determination of the optimal shape of a surface electrode: Selective stimulation. *Journal of Neuroscience Methods* 178, 1 (2009), 174 – 181.
29. Prausnitz, M. R. The effects of electric current applied to skin: a review for transdermal drug delivery. *Advanced Drug Delivery Reviews* 18, 3 (1996), 395–425.
30. Prutchi, D., and Norris, M. *Design and development of medical electronic instrumentation: a practical perspective of the design, construction, and test of medical devices*. John Wiley & Sons, 2005.
31. Tamaki, E., Chan, T., and Iwasaki, K. UnlimitedHand: Input and output hand gestures with less calibration time. In *Proceedings of the 29th Annual Symposium on User Interface Software and Technology*, ACM (2016), 163–165.

32. Tamaki, E., Miyaki, T., and Rekimoto, J. PossessedHand: techniques for controlling human hands using electrical muscles stimuli. In *Proceedings of the SIGCHI Conference on Human Factors in Computing Systems*, ACM (2011), 543–552.
33. Tuckett, R. P. Itch evoked by electrical stimulation of the skin. *Journal of Investigative Dermatology* 79, 6 (1982), 368–373.