DextrEMS: Increasing Dexterity in Electrical Muscle Stimulation by Combining it with Brakes

Romain Nith
University of Chicago
rnith@uchicago.edu

Shan-Yuan Teng
University of Chicago
tengshanyuan@uchicago.edu

Pengyu Li
University of Chicago
pengyulee@uchicago.edu

Yujie Tao
University of Chicago
yujiet@uchicago.edu

Pedro Lopes
University of Chicago
pedrolopes@uchicago.edu

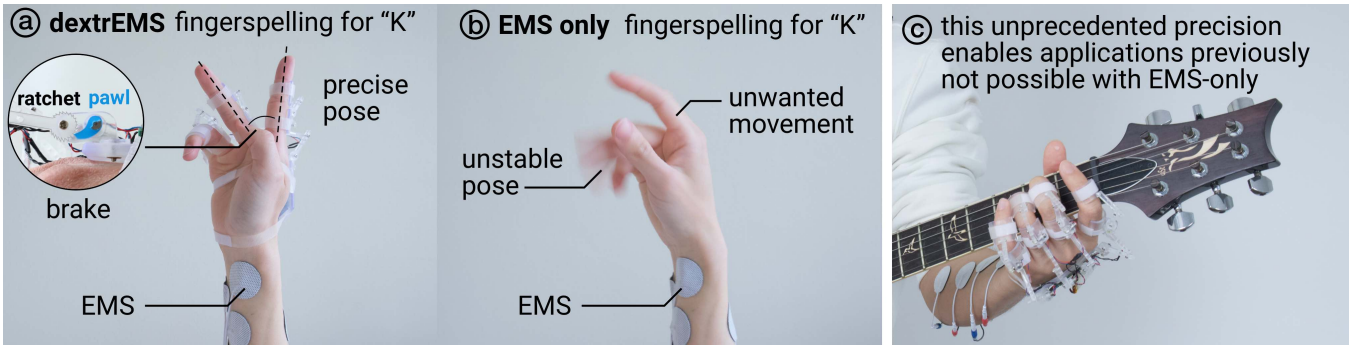

Figure 1: DextrEMS is a haptic device designed to improve the *dexterity* of electrical muscle stimulation (EMS). It achieves this by combining EMS with a mechanical brake on all finger joints. These breaks allow us to solve two fundamental problems with current EMS devices: (1) lack of independent actuation (i.e., when a target finger is actuated via EMS, it also often causes unwanted movements in other fingers); and (2) unwanted oscillations (i.e., EMS cannot stop a finger at a precise angle without oscillations, which originate from the fact that to stop a finger, EMS needs to continuously contract the opposing muscle). Using its brakes, dextrEMS achieves unprecedented dexterity, in both EMS finger flexion and extension, enabling applications not possible with existing EMS-based interactive devices. For instance, (a) we demonstrate a haptic finger spelling application, that actuates the user's fingers to pose the "K" sign (in American sign language, from which dextrEMS can only render a very small subset); (b) the quality of the same pose via EMS alone suffers from oscillations and unwanted movements; or, lastly, (c) a haptic guitar tutorial that actuates the user's fingers to form an E-minor chord.

ABSTRACT

Electrical muscle stimulation (EMS) is an emergent technique that miniaturizes force feedback, especially popular for untethered haptic devices, such as mobile gaming, VR, or AR. However, the actuation displayed by interactive systems based on EMS is coarse and imprecise. EMS systems mostly focus on inducing movements in large muscle groups such as legs, arms, and wrists; whereas individual finger poses, which would be required, for example, to actuate a user's fingers to fingerspell even the simplest letters in sign language, *are not possible*. The lack of dexterity in EMS stems from two fundamental limitations: **(1) *lack of independence***: when a particular finger is actuated by EMS, the current runs through nearby muscles, causing unwanted actuation of adjacent fingers; and, **(2) *unwanted oscillations***: while it is relatively easy for EMS to start moving a finger, it is very hard for EMS to *stop and hold* that finger at a *precise angle*; because, to stop a finger, virtually all EMS systems contract the opposing muscle, typically achieved via controllers (e.g., PID)—unfortunately, even with the best controller tuning, this often results in unwanted oscillations. To tackle these limitations, we propose dextrEMS, an EMS-based haptic device featuring mechanical brakes attached to each finger joint. The key idea behind dextrEMS is that while the EMS actuates the fingers, it is our mechanical brake that *stops* the finger in a precise position. Moreover, it is also the brakes that allow dextrEMS to select which fingers are moved by EMS, eliminating unwanted movements by preventing adjacent fingers from moving. We implemented dextrEMS as an untethered haptic device, weighing only 68g, that actuates eight finger joints independently (metacarpophalangeal and proximal interphalangeal joints for four fingers), which we demonstrate in a wide range of haptic applications, such as assisted fingerspelling,

UIST '21, October 10–14, 2021, Virtual Event, USA

ACM ISBN 978-1-4503-8635-7/21/10…$15.00
https://doi.org/10.1145/3472749.3474759

a piano tutorial, guitar tutorial, and a VR game. Finally, in our technical evaluation, we found that dextrEMS outperformed EMS alone by **doubling its independence and reducing unwanted oscillations**.

KEYWORDS

electrical muscle stimulation, exoskeleton, dexterity, force feedback, haptics

ACM Reference Format:
Romain Nith, Shan-Yuan Teng, Pengyu Li, Yujie Tao, and Pedro Lopes. 2021. DextrEMS: Increasing Dexterity in Electrical Muscle Stimulation by Combining it with Brakes. In *The 34th Annual ACM Symposium on User Interface Software and Technology (UIST '21), October 10–14, 2021, Virtual Event, USA.* ACM, New York, NY, USA, 17 pages. https://doi.org/10.1145/3472749.3474759

1 INTRODUCTION

EMS is a popular technique for force-feedback and haptic-actuation because of its small form factor when compared to mechanical actuators (e.g., exoskeletons [75]). As such, non-implanted EMS has been increasingly used to replace traditional mechanical force-feedback devices and enable a wide range of untethered/mobile applications, such as: force-feedback in virtual [21, 52] or augmented reality [53] (VR/AR), moving a user's wrist to tap to a beat [18, 19], teaching users how to manipulate unfamiliar objects [50], or mobile information access [49].

However, all these aforementioned non-implanted EMS systems, and all others in the literature, are limited in that they either do not actuate the user's fingers (instead they actuate larger muscle groups, e.g., arms [30, 52], wrists [47, 50, 52–54] or legs [66]) or actuate the user's fingers very *coarsely* and not *independently* of each other; in other words, current interactive systems based on EMS display two key issues: **(1) lack of independent finger actuation**: when a particular target finger is actuated by EMS, the other fingers are actuated as well, causing a lack of *dexterity*—this explains why most authors only envision EMS to enable dexterous applications, such as for playing guitar, but we have yet to see any interactive EMS application realized with this level of finger independence; and, **(2) unwanted oscillations**: while it is relatively easy for EMS to start actuating the user's finger muscles, it is very hard for EMS *stop and hold* one particular finger at a *precise angle/pose*, because to prevent a finger from moving, virtually all EMS systems continuously contract the opposing muscle, typically relying on a PID controller that regulates this isometric muscle contraction [37, 43, 49, 72, 74, 88]—unfortunately, even the best tuning still results in unwanted oscillations as the controller stops the finger at the target pose, these oscillations are detrimental to the user experience as they send unnecessary proprioceptive signals to the user.

This lack of dexterity in EMS-induced finger movements was well documented in the *PossessedHand* [84], which pioneered the use of EMS to engineer interactive devices. In this system, the authors found that they could only control five out of 16 joints independently of the other fingers; in other words, 11 out of 15 joints moved together (11 joints with unwanted movements). Although 10 years have passed since the *PossessedHand,* it still stands as the most dexterous finger actuation in all interactive EMS devices. While optimizing electrode layouts can minimize some of the unwanted movements that limit EMS' independence [82], this has only been applied to *full flexions* around the metacarpophalangeal joints (MCP), while proximal interphalangeal joints (PIP) and extensions of any fingers were not considered; as a result, it can only flex the full finger but cannot *hold it* at any precise angle.

The root of non-implanted EMS's lack of dexterity stems from: (1) forearm muscles that control fingers are *densely packed*, as they all meet at the elbow with the humerus bone as a shared anchor; therefore, stimulating *one* of these muscles by electrodes attached to the skin causes currents to run also via adjacent muscles, causing other muscles contract *unwantedly*; and, (2) forearm muscles that control movement around the different joints of the same finger (e.g., MCP vs. PIP) are *layered*, with muscles that control MCP movement at one depth (*flexor digitorum profundus*) and muscles that control PIP at *multiple* depths (jointly *digitorum superficialis* and *digitorum profundus*); as such, stimulating a finger around a particular joint (e.g., MCP) by electrodes attached to the skin usually causes currents to run through other layers, resulting in unwanted movements in other joints (e.g., PIP joint or even the wrist).

To illustrate the importance of this problem, we found that, in just the last 17 years of EMS research in HCI/Haptics, 54 publications [1, 7, 11, 13–19, 21, 23, 24, 26, 28, 30, 33–41, 43, 45, 47, 50, 54, 55, 61–65, 67–71, 77–79, 83–89] use EMS for finger movements but do not exhibit any more dexterity in the resulting finger movements than the *PossessedHand,* neither they surpassed these two key limitations.

In this paper, we demonstrate a haptic device that increases the dexterity of EMS-based finger actuation. Our approach, which we call *dextrEMS* (a contraction of dexterous-EMS), is a haptic device that actuates eight finger joints (MCP and PIP) providing an unprecedented performance compared to existing non-implanted EMS devices. It addresses the limitations of EMS by combining it with a mechanical brake at every finger joint. This brake (1) holds the fingers when they reach a desired target angle/pose; and (2) prevents adjacent fingers from moving, by locking fingers not involved in the target pose. Moreover, because our brake mechanism (a custom-made ratchet) is *only* used to halt the fingers and *not to actuate them,* the resulting device is still lightweight (68g) for mobile applications.

2 DEXTREMS'S KEY PRINCIPLE: EMS + BRAKE = PRECISE PER-FINGER ACTUATION

To tackle the aforementioned limitations that prevent EMS from dexterity, we propose combining each finger moved by EMS with a *brake added at every joint*, which is depicted in Figure 2 (b). We demonstrate this principle at the example of tackling dexterous EMS finger actuation, a long-sought challenge in EMS. Applying our principle to the fingers results in a compact wearable device, which is depicted in Figure 2 (a), that generates force feedback with more dexterity (i.e., the quality of the resulting pose) than existing EMS devices.

The key principle behind dextrEMS is depicted in Figure 3 at the example of fingerspelling the letter "K" in American Sign Language (ASL); it is important to note that dextrEMS does not solve any

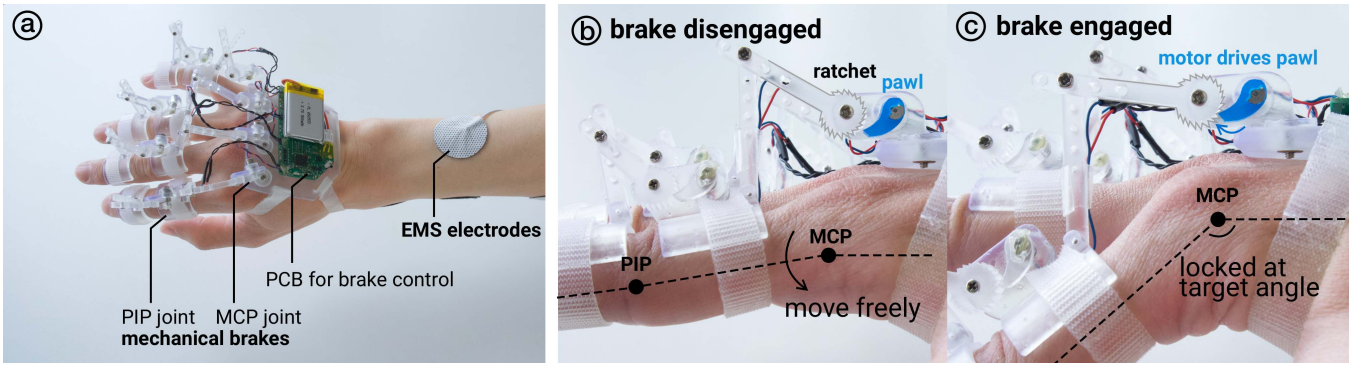

Figure 2: (a) DextrEMS is stand-alone via a custom PCB that drives the brakes, battery, and wireless. EMS electrodes are placed on the dorsal side for finger extension, and on the palmar side for flexion. (b) Our brake mechanism. (c) We use one ratchet per MCP and PIP joint on all four fingers, which allows us to stop the finger without the oscillations observed in typical EMS control loops.

inherent challenge faced by the hard of hearing or deaf community, not only because of its limitations (only renders a small subset of fingerspelled letters, mostly simple poses that do not cross fingers or use thumb grips) but, more importantly, because no technological solution should "just solve" the unique experience of these individuals. Rather, we were inspired by the ASL alphabet and use it to demonstrate the dexterity of dextrEMS when compared to only using EMS.

Figure 3 decomposes dextrEMS' actuation into three phases: (a) brakes lock joints that are not meant to move; (b) EMS actuates the fingers, but its lack of independence is not experienced as our brakes prevent unwanted movements in non-target fingers; finally, (c) when each finger reaches the fingerspelling "K" pose (measured by an external hand-tracking system) the mechanical brakes lock and *halt* the finger—this results in hand poses without the typical oscillations seen as EMS-only systems attempt to halt a limb in a precise position (e.g., [37, 49]).

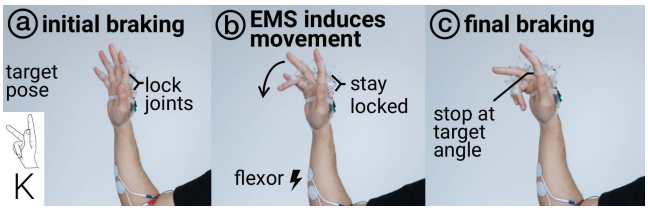

Figure 3: Working principle behind our haptic device. In this example, the target pose is the letter "K" in American Sign Language. (a) Braking mechanism locks finger joints that aren't supposed to move. (b) EMS on the flexor actuates the fingers, bringing down free-moving finger joints. (c) Brakes on the moving fingers lock when the target angle is achieved, sensed with an external camera, stopping the finger in a steady state. Our device allows posing without complex EMS control loops and less EMS stimulation.

Moreover, because our brake mechanism is implemented using a small ratchet and a matching pawl, it only requires a very small motor to control the brake (0.13A at 4.2V per brake for only 12.5ms) while only weighing 68g. This is a stark contrast with the typical size, weight, and power consumption of active exoskeletons, which require powerful and heavy motors with sufficient force to physically stop the finger, but, more importantly, it provides an *improvement in dexterity* when compared to the dexterity offered by using EMS alone.

This unique combination of EMS and mechanical brakes is the key to allowing dextrEMS to improve EMS' dexterity. To better illustrate this, in Figure 4 we depict the same pose from Figure 3 but using EMS alone. Here, to fingerspell the "K" pose using EMS alone, it first stimulates the finger flexors that control the pinky, ring, and middle finger. Unfortunately, these muscles are adjacent to one another inside the forearm, and EMS currents run through adjacent muscles causing unwanted movements, which is depicted in Figure 4 (b). More dramatically, while it is relatively easy for EMS to start moving a finger, it is very hard for EMS to stop and hold that finger at a precise angle; because, to stop a finger, virtually all EMS systems contract the opposing muscle, typically achieved via controllers (e.g., here we use a dual-sided PID that stimulates both flexors and extensors)—unfortunately, even with best controller tuning, this still often results in unwanted oscillations, as depicted in Figure 4 (c).

Figure 4: (a) Traditional EMS device actuates the fingers with groups because of layered muscles under the skin, resulting in moving unwanted finger joints. (b) EMS devices utilize actuation of the opposing muscle group to attempt at slowing down/stopping the moving limb, this often results in (c) an unstable pose that oscillates around the target.

3 BENEFITS, CONTRIBUTIONS, AND LIMITATIONS

The key contribution of our work is a new way to minimize two of the key limitations of current interactive devices based on EMS; we achieve this by contributing a new haptic device that combines EMS with mechanical brakes.

The benefit is that this results in: (1) higher dexterity than current EMS-based haptic devices in both EMS finger flexions and extensions. Our device doubles the independence of EMS (i.e., it can move a specific finger without unwanted movement of other fingers) and reduces the amount of unwanted oscillations; (2) dextrEMS enables new interactive applications not possible before with EMS, for instance, fingerspelling simple letters in ASL, or actuating a user's hand to form a chord on a guitar's fretboard. (3) When compared to fully actuated exoskeletons, which tend to have more dexterity than EMS alone, our device is lighter because our brakes are based on a ratchet, which locks mechanically and not by forcing against the user.

Our haptic device is not without limitations: (1) like any other device based on EMS it requires electrode placement and calibration. Moreover, like in other EMS devices with electrodes at the forearm, turning the wrist might also break the alignment between the electrodes and the muscles, hindering the actuation; (2) as any other passive exoskeleton, it covers part of the user's hand; note we intentionally designed our mechanics as such to minimize this interference; in fact, unlike most exoskeletons, we ensured that the user's fingerpads are not covered in any way to enable maximum tactile feedback (e.g., one can comfortably type on a mobile phone while wearing dextrEMS); (3) while it controls flexions and extension of all four fingers around both MCP and PIP joints (eight per hand in total, two directions, thus 16DoF), it does not actuate the distal interphalangeal joints nor the thumb; lastly, (4) while our device doubles the independence of finger actuation when compared with EMS alone, it is limited by the resolution of the ratchet that implements the brake. Our laser-cut ratchet has 24 teeth, which results in a brake position every 15°. One can simply double (or even more) the resolution of dextrEMS by CNC-machining the ratchet out of stronger materials (e.g., metal rather than acrylic).

4 RELATED WORK

The work presented in this paper builds primarily on the field of wearable haptics, in particular exoskeletons and electrical muscle stimulation.

4.1 Towards wearable force-feedback devices

Force-feedback devices apply force on the user's body to create sensations of weight or resistance or even actuate their limbs. Typically, force-feedback devices require substantial power and are thus constructed from grounded motors, e.g., *SpidarG&G* [60], *EXO-UL7* [73], or *SensAble PHANTOM Omni* [59]. With much of today's interactions taking place anywhere and anytime, i.e., mobile settings, researchers have been exploring how to miniaturize force-feedback devices. The most popular approach is to make devices wearable, i.e., attach an actuator to the user and push against the user's body—this is called an *exoskeleton* [3, 5, 6, 25, 29, 57, 73, 81].

4.2 Wearable force feedback using motors (active exoskeletons) vs. brakes (braking exoskeletons)

Using glove-like exoskeletons worn on the user's hand provides force feedback to the fingers, which is a popular approach in rehabilitation, haptic guidance, and feedback for immersive experiences. Exoskeleton gloves enjoyed this popularity for a number of decades, for instance, the *CyberGrasp* [12], *Haptic Telexistence* [76], and *RML Glove* [57] are canonical examples that used strong motors that can actuate the user's limbs to provide force feedback. These exoskeletons create force feedback on the fingers by *actively* resisting their movement using motors, which in order to provide sufficient force to stop a finger, turn out to be large. As such, their final form factor is prohibitively large, heavy, and often not standalone, e.g., power supplies are external.

As an alternative to this, researchers turned to *brake-based* exoskeletons, i.e., these use very small motors coupled with passive brakes, utilizing the fact that the resisting limb movements are realized passively through brake mechanism and not by motors pushing against the limb. *Wolverine* [10] and *Grability* [9] are canonical examples that demonstrate how unidirectional brakes between three fingers and a thumb can provide force feedback in VR. *Dexmo* [25] is a lightweight exoskeleton using small servos combined with rachets and linkages that can stop the finger movement interactively.

Other alternative mechanisms to resist finger movement also exist, such as particle jamming using external vacuum pumps (e.g., in *Jamming glove* [92]), pistons [5] [20], layer jamming [8], electrostatic brakes [31], or even magnetorheological fluids [4]. All these alternatives can generate high braking force with the help of large equipment (high voltage supply or external pumps) or have diminished performance when scaled down. These factors make them unsuitable for portable or wearable applications. We are inspired by these previous works and designed a braking system combined with EMS that does not require any heavy or power-hungry components—our brake mechanism uses only 0.13A for 12.5ms to halt a moving finger and is made from lightweight acrylic.

4.3 Electrical Muscle Stimulation

Electrical muscle stimulation (EMS) is a more recent alternative to the long-standing challenge of actuating limbs. EMS involves solely attaching electrodes to the user's skin, atop a muscle. Then, passing a small current through those electrodes causes the muscle fibers to contract and, in turn, actuate the user's muscle. As such, many hail EMS as an increasingly promising technique to miniaturize strong force feedback at a very small form factor [46].

EMS stems from a long history of developments in medical rehabilitation, back to the 1960s [80]. There, it has been one of the primary techniques for restoring lost muscular functions (e.g., often as a result of spinal cord injury [80], stroke [32], or drop-foot syndrome [91]).

Only more recently, researchers explore the idea that EMS can replace mechanical force feedback actuators [46]. Since then, EMS has been permeating interactive devices, since not only does it affords force feedback but it accomplishes this at an especially lightweight form-factor [30, 47–50, 52–54, 66] much needed to build small/wearable haptic devices.

However, EMS suffers from limitations, one of the most striking being: **its lack of dexterity**. This lack of dexterity is exacerbated by the fact that to interactively stimulate the muscles of a user with EMS, researchers typically put the electrodes *on top of the skin* (not implanted). As such, the electrodes cannot access a muscle precisely without also *letting currents pass through adjacent muscles* or even *different depths*. These limitations have been documented in the early HCI works that pioneered the use of EMS in interactive devices, for instance: "This contraction is hard to control" (from Kruijff et al. [42]) and, similarly, "We confirmed that PossessedHand could control 5 independent and **11 linked joints** [11 out of 16 joints have unwanted movements]" (from PossessedHand [84]).

Because of this limitation, it is difficult to individually control each finger movement with dexterity. As such, most of the work using EMS focuses on coarse body movement such as on the wrists [47, 49, 50, 52–54], arms [30, 52], and legs [66]. The applications of EMS range from eyes-free interactions [49], mobile gaming [47], virtual [21, 52], or augmented reality [53], to augmenting object affordance in daily life [50]. Our work aims to leverage the small form factor of EMS and improve its dexterity in terms of finger pose by using a lightweight exoskeleton.

4.4 Adding control loops to stabilize the movement induced by EMS-based actuators

To combat the difficulty of accurately controlling human limbs through EMS, researchers have employed closed-loop controllers, especially the proportional-integral-derivative controller (known as PID), which regulates the EMS applied to opposing muscles so as to stop the moving limb at a target angle [43, 49, 72, 74]. To better illustrate how EMS-based interactive systems suffer from oscillations typical of PID controllers, we look at two examples: (1) Kaul et al. [37] added a PID controller to their EMS system that actuated the user's arm to point a target, but found overshooting often affected the trajectories in their study; similarly, (2) Watanabe et al. [88] added a PID control to their EMS system to actuate the MCP and PIP joints of the user's middle finger, but found oscillations occurred when reaching the target joint angle. As we can observe from prior work, *even the best tuning by these expert researchers* still resulted in unwanted oscillations as the controller attempts to stop the limb at the target pose. These oscillations are detrimental to the user experience as they send unnecessary tactile and proprioceptive signals to the user. Note that while PID tends to be the chosen controller by most researchers, likely due to its capabilities for handling over/under-shooting, others have employed alternative methods such as an extended Kalman filter used in Widjaja et al. [90].

Moreover, other advances in EMS are expected to improve the quality of EMS' dexterity, such as optimizing the placement of electrodes that control finger muscles [1], automatic calibration of high-density electrode arrays [39], or even, new electrode layouts that offer some minimization of the unwanted actuation [82]. However, the latter only allows to achieve a *full flexion* around the MCP (no PIP joint, nor extensions of any other fingers); in other words, it can only flex the full finger at the MCP joint but not *stop it* at any precise angle along the way. As such, the challenge of achieving EMS dexterity without oscillations remains, which is the focus of our device.

5 IMPLEMENTATION

To help readers replicate our design, we now provide the necessary technical details. Furthermore, to accelerate replication, we provide all the source code, firmware, 3D files, and schematics of our implementation[1].

DextrEMS is implemented as a self-contained haptic device (i.e., it works on battery and is wireless) comprised of two key components: (1) a custom lightweight glove with eight brake-based joints and its controller electronics of a total weight of 68g; and (2) any existing EMS stimulator with at least four channels. Our haptic device is secured to the user's hand with Velcro straps.

5.1 Mechanics: braking system

The majority of our brake mechanism was 3D printed (Form Labs 3) using clear resin, while the hinges at each joint were laser cut out of 3mm clear acrylic. The see-through materials were used to minimize visual obstruction of the real world.

The mechanism of one of our brakes, depicted in detail in Figure 2, is a custom-made ratchet and pawl mechanism. This is a standard mechanism that allows for rotary motion in only one direction because in the opposite direction the pawl jams against the depression between the ratchet's teeth. Unlike passive ratchet and pawl mechanisms (as found in many everyday tools, such as wrenches) our pawl is controlled by a small DC motor (Vibration Motor 11.6×4.6×4.8mm, Polulu). To activate the brake, the dextrEMS circuit drives the DC motor clockwise, which jams the pawl into the ratchet, or counterclockwise to release it from the ratchet. Like any device based on a ratchet mechanism, our brake is discrete because the ratchet has a fixed number of teeth that dictate the final resolution of the brake. Our laser cut ratchet has 24 teeth, which results in a brake position every 15°; while this is the precision limit of our current implementation, to surpass it, one only needs to produce a new ratchet; for instance, a CNC machined ratchet can feature the double (or more) teeth in the same form factor as our laser cut one.

While our ratchet and pawl combination does not implement bidirectional mechanics, dextrEMS does achieve bidirectional braking by leveraging EMS and the biomechanics of the fingers to brake in both the flexion direction (the natural brake direction of our ratchet) as well as in the extension direction. To brake in the latter direction, we preemptively actuate the pawl, earlier than in the flexion direction, i.e., we actuate the pawl one brake position earlier. Because the finger is moving in the extensor direction, this does not immediately cause the pawl to jam in the ratchet. Thus, we subsequently, stop the actuation in the extensor direction, causing the finger the recoil back to the resting pose, which anatomically is below extension (neutral, towards flexor). This creates a momentary flexion as the finger recoils back, which now actuates the pawl in the correct direction to jam inside the ratchet—achieving braking also in the extensor direction.

Our mechanics are not without their limitations. Any exoskeleton will exhibit some mechanical compliance (i.e., the so-called "slack"). To this end, we used acrylic, a stiff material, in our design to mitigate some of this detrimental compliance. Furthermore, we

[1] https://lab.plopes.org/#dextrEMS.

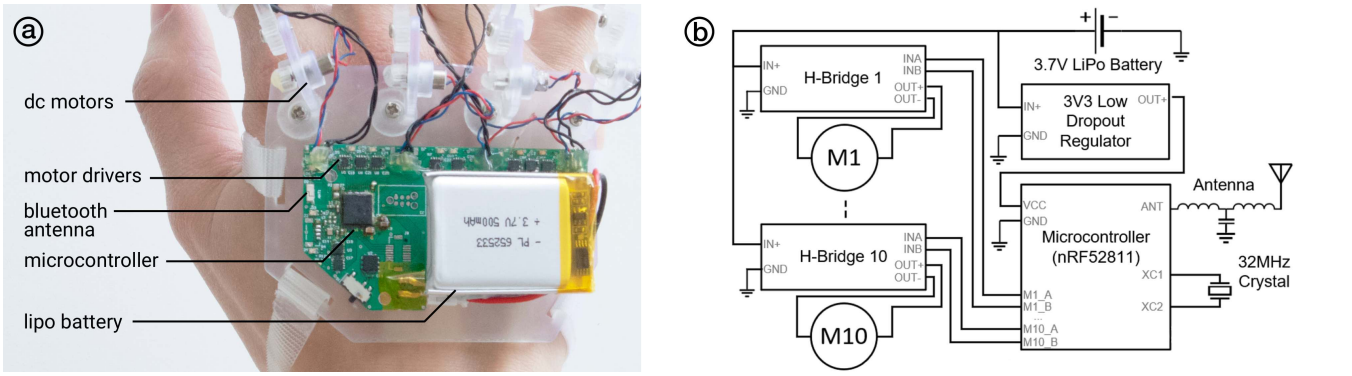

Figure 5: (a) Final self-contained printed circuit board. (b) Electronics schematic of our haptic device.

added two Velcro straps along the metacarpal bone, as well as additional straps for the middle and proximal phalanx of each finger, to secure the exoskeleton to the hand. These straps ensured an average 105mm^2 of contact between the hook and loop, which reduced mechanical slack. As such, after controlling for these variables, we believe the main limitation of the mechanical resolution stems from the aforementioned ratchet teeth count. Again, fabricating a ratchet with a higher resolution (e.g., from metal) will likely increase the accuracy.

Finally, as with most exoskeletons, the fit is never universal. However, dextrEMS accommodates various hand sizes by adjusting the Velcro straps as well as our length-tunable hinges, which provide several pin-holes to adjust the length of each joint, as depicted in Figure 2 (b, c).

5.2 Electronics: circuit design and printed circuit board

The electronics of our device, depicted in Figure 5, are housed in our custom printed circuit board. Fitted on the back of the hand, its core is an nRF52811 microcontroller (Nordic Semiconductor) with on-chip Bluetooth Low Energy (BLE). Each of our eight 11.6×4.6×4.8mm DC motors (Polulu Vibration Motor, one placed at every PIP and MCP joints in all four fingers) is driven by a DRV8837 H-bridge motor driver (Texas Instruments). Moreover, to better visualize the action of each brake joint, we added a red or green LED parallel to the positive and negative rail output of the motor-driver, allowing us to visually see if a brake is locked or unlocked (refer to our Video figure).

The electronics are all powered via a 500mAh 3.7V LiPo battery. We measured a peak current draw of 1.02A when all eight motors are stalling, which occurs when all pawls are being moved into their respective ratchets. While this depicts the highest current draw possible (all joints need to be locked at the same time) it is extremely brief (~12.5ms) and quite low-powered since as soon as the pawl is jammed into the ratchet, the motor drivers release the DC motor, relying on the *mechanical* force to keep the lock. This solution enables our device to be power efficient, especially when compared to larger exoskeletons built using much larger DC or servo motors that can consume much more than 1A per individual motor [6, 22, 57].

5.3 Electrical muscle stimulation

The EMS stimulator we employ in dextrEMS is derived from previous EMS research, demonstrating its applicability to improve the dexterity of *existing* EMS systems. It implements a unidirectional PID controller, tuned per joint (similar to [49]). The PID controller takes as input the angles of each finger joint and outputs the pulse-width of the EMS signal required to achieve this angle. Like any EMS-based PID controller, its key contribution is to slow down the finger's inertia as it is approaching the target to avoid any brake overshoot.

Using EMS, we address four muscle groups of the forearm, in particular: (1) *flexor digitorum superficialis* at the location where it flexes predominately the index finger (MCP and PIP); (2) *flexor digitorum superficialis* at the location where it flexes predominately the ring finger and middle finger (MCP and PIP); (3) *flexor digitorum superficialis* at the location where it flexes pinky finger (MCP and PIP); and, (4) *extensor digitorum*, which extends all four fingers (MCP and PIP); all these placements are inspired to those used in PossessedHand [84].

To actuate the fingers by means of EMS, we use a Rehastim3 medically-compliant muscle stimulator, this device is battery powered and communicates with dextrEMS using an additional host computer with USB and BLE. Then, to enable mobile use, we modified the Rehastim3 library [51] to compile on ARM V7+ architecture (which we provide in[1] for replication and the benefit other researchers), allowing us to control the EMS device and dextrEMS from the 55.0mm x 35.0mm microprocessor from the *PocketBeagle* [2].

5.4 Tracking and haptic communication pipeline

Tracking. While our haptic device is self-contained, using dextrEMS requires finger tracking, which is typically supplied by the remainder interactive apparatus (e.g., VR or AR headsets, and so forth). In all our demos we leveraged existing tracking systems. In particular, in our VR applications, we leveraged the built-in finger tracking from the *Oculus Quest*, while in our mobile-phone applications, we used *MediaPipe* finger tracking [56]. Later, we will also demonstrate an alternative tracking system that we integrated during the early stages of our device, which was based on flex sensors;

however, we found the latter to be less robust than the remainder optical tracking systems.

Haptic communication pipeline. To illustrate our haptic pipeline, we describe the steps involved in actuating a finger in our VR piano tutorial: when the VR application intends to move the user's fingers to strike a note of the piano, it sends a "actuate" message to dextrEMS over BLE, which indicates that flexion is needed on the index finger. DextrEMS responds by locking all other fingers and actuating the EMS channel corresponding to the index and middle finger. Then, the VR application tracks the finger's movement, using its built-in Oculus Quest hand tracking and monitors for any collision between the finger and any 3D object in the scene. As the finger pushes the piano key to its final position, the VR application sends a "stop" message to dextrEMS, indicating to stop the index finger. DextrEMS responds by locking the index finger and disabling the EMS stimulation. Finally, our smartphone application (*Android* or *iOS)*, such as the haptic fingerspelling application, use a similar haptic pipeline but use the front-facing camera of the phone and MediaPipe as their tracking system. To enable the detection of simple sign language gestures, we pre-trained on a small set of targeted (ASL) gestures. The tracking and gesture recognition runs in real-time.

5.5 Alternative mechanical designs: (a) sideways PIP brake; (b) thumb-mechanism; (c) flex-sensors

While working on our implementation we also engineered three alternatives depicted in Figure 6 (a) a brake for the PIP joint that did not use hinges that extrude vertically above the fingers; (b) a preliminary brake mechanism for the thumb; and, (d) a simple integrated finger tracking system using flex sensors. We present these mechanical designs as we believe they will assist future researchers in creating variations of dextrEMS for applications we did not explore.

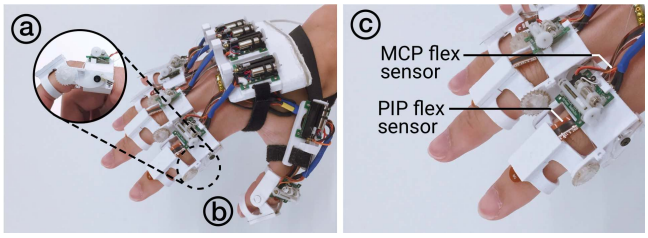

Figure 6: The alternative mechanical designs we also explored while engineering our device: (a) a sideways brake, mounted directly at the PIP joint, allowing for a smaller vertical form factor but restricting the lateral finger movement (finger adduction); (b) one preliminary design for the thumb joint, which allows PIP locking but only weak MCP locking; and, (c) integrated sensing using simple flex sensors at every joint.

Sideways PIP Brake. We implemented an alternative mechanical design for the PIP joint that did not use vertical hinges. Instead, we implemented a ratchet and pawl configuration that was positioned laterally and, thus, directly at the PIP finger joints, as depicted in Figure 6 (a). This alternative mechanism results in a slimmer vertical form factor but wider horizontal form factor; as such, it is harder to close the fingers (adduction), which is the reason why we opted for our vertical hinge design that affords much more dexterity. Moreover, this mechanical design cannot be applied to the MCP joint, so it is only suited for applications that make use exclusively of PIP-actuation and do not require laterally closing the fingers together. This alternative PIP brake follows a similar mechanical principle as our main design showcased in the Implementation section. However, this mechanical design cannot be directly driven because the finger itself is in the way. As such, it uses a pulley-based redirection to brake: the pawl jams into the ratchet as it is pulled by a wire attached to a small 1.8g linear servo motor (SPMSA2005, Spektrum). To retract, we spring-loaded the mechanism using a spring at the back of the pawl, which keeps it disengaged when the linear servo motor loosens up the pulley.

Exploring a brake for the thumb. We also engineered a simple brake mechanism for the thumb based on our sideways PIP joint and a simple pulley for the MCP joint. However, this pulley is not as effective as the ratchet-pawl brake because it does not lock, i.e., it relies on force rather than on a locking mechanism. This approach proved only sufficient for applications that do not require strong forces on the thumb's MCP. As such, we decided to not implement this mechanism on our final device. Finally, it is worth noting that the thumb has an additional degree of freedom, achieved by an additional joint: the carpometacarpal (CMC) joint. This joint was not targeted by our brake system.

Integrating an on-board finger tracking using flex sensors. While our main dextrEMS device relies on finger tracking supplied by the interactive applications (e.g., VR applications on Oculus Quest or phone applications that track via MediaPipe), we also explored a simple integrated finger tracking system using flex sensors, which is depicted in Figure 6 (c). To implement this, we instrumented each MCP and PIP with two bend sensors, which tracked the flexion of each joint. To compensate for the change in diameter when the bend angle increases, the sensors are anchored on one side of the exoskeleton and are freely sliding on the opposite side, as depicted in Figure 6 (c).

6 TECHNICAL EVALUATION

The goal of our technical evaluation was to compare the performance of dextrEMS to actuating the finger flexor/extensor muscles using EMS alone. As such, we designed a test setup that allowed us to measure the two key properties that we expected our device to improve: **(1) finger MCP and PIP independence**, i.e., to which degree can a haptic device actuate a finger around the MCP and PIP joints independently of other fingers; and, (2) **pose precision vs. unwanted oscillations**, i.e., to which degree of precision can a haptic device pose a target finger to match a specific angle without unwanted oscillations. Finally, to provide more insights into the applicability of our system we also measured its: **(3) end-to-end latency,** and **(4) maximum braking force**. To assist the reader with replicating our technical evaluation, we provide all source code[2] (EMS/dextrEMS controllers, *Optitrack* script, analysis scripts, and our custom-made image labeling software).

[2]https://lab.plopes.org/#dextrEMS

Apparatus. We constructed a simple experimental apparatus, comprised of a stand that secures the participant's arm, allowing us to stabilize the elbow and forearm. For our evaluations, we recruited a participant per study, from our local institution, with no previous experience with our device. The participant's reactions to the interface were not the focus of our experiments nor do we aim to generalize beyond it. Instead, this technical evaluation was designed to measure how our braking mechanism improves the dexterity of EMS alone.

6.1 Technical evaluation#1: Finger independence (i.e., moving each finger's MCP & PIP joint without other fingers)

Interface conditions. In this technical evaluation, we measured finger independence for MCP and PIP flexion, comparing: **(1) our dextrEMS** device, connected directly via USB; and, (2) the **existing EMS** approach, inspired by the description in PossessedHand [84]. It is important to note that the dextrEMS condition uses precisely the same EMS setup as the baseline (calibrated the same way), the only difference being that the participant is additionally wearing our braking-exoskeleton in combination with EMS. The goal was to flex each finger (index, middle, ring, and pinky) at a target joint (MCP or PIP) and measure how much the other fingers moved unwantedly.

Apparatus. To obtain the angles of all the MCP and PIP finger joints, the participant's hand was filmed from two angles by two cameras at 30fps. Camera images were corrected with checkerboard-method for lens distortion. This allowed us to precisely measure all the angles of all the MCP and PIP joints by manually labeling each image using a custom script implemented using *OpenCV*. Note that prior to deciding on a camera-based approach, we compared this and our Optitrack motion tracking system or, even, placing MPU9250 9DOF inertial measurement units (IMUs) with Madgwick filters [58] on all joints. We compared the angle estimation of each tracking system against a ground truth protractor and found that the camera approach resulted consistently in the highest precision.

Calibration. We calibrated the EMS for each participant., per-finger joint. First, to determine the stimulation intensity (i.e., current, in mA) we started with an intensity of 0mA and a pulse-width of 300 μs and slowly increased the intensity in 1 mA steps until the participant's finger was fully flexed while minimizing any additional movement on adjacent fingers; then, if necessary to achieve sub-mA adjustments, we fixed the intensity and subsequently adjusted the pulse-width. We repeated this for all joints of the four fingers, essentially calibrating following a manual version of PossessedHand [84]: (1) attached electrodes to the target muscles following anatomical guide; (2) per electrode, started with low pulse-width and increased it step-by-step, confirming at each step with the participant that the stimulation was is pain-free; (3) repeated until no more flexion happens at the target finger or pain has been reached (the latter never occurred). Finally, it is worth noting that we have years of experience with this EMS calibration that follows PossessedHand's [84].

Procedure. In this test, we recorded a total of 48 trials: four fingers (index, middle, ring, and pinky) x two joints (MCP and PIP) x three repetitions x two interface conditions. Per trial, the EMS actuated one target finger and the video cameras recorded the movement. The order of the interface conditions and the order of the finger actuation was randomized. Once all fingers had been actuated, we moved onto the second interface and repeated for all four fingers in a newly randomized order. As a result, each trial depicts the final pose from the resulting actuation.

Accuracy metric: independence index. Our study uses the independence index (also known as I.I. in hand kinematics literature), a standard metric used to measure the amount of independent movement of a finger. This index denotes the ratio of movement between a finger's joint (measured in the angle of the joint's movement) to how much the other joints moved [44]. Thus, when considering both the MCP and PIP joints, the independent index of a target finger i (II_i) can be calculated as follows:

$$II_i = 1 - \frac{\frac{1}{7}\sum_{k\in\Gamma_i}|A_k|}{|A_i|}, \qquad \text{where,} \quad \mathrm{A_k} = \int_{\mathrm{T}} \theta_k \mathrm{d}t,$$

Here, i denotes a stimulated target finger and Γ_i is a set of not-target fingers. An independence index of 0 or lower indicates that actuating the target finger around this joint caused movement in the other joints (including even joints in the same finger). Conversely, an index closer to 1 indicates more independence, with a perfect "1" depicting a complete independent movement of the target finger around this joint (with no unwanted movements from other joints). However, it is critical to note that, anatomically, the human hand does not exhibit fully independent fingers. As experienced in daily life, many fingers bend when other fingers move. Lang et al. measured an independence index of 0.84 for passive finger flexions around the MCP joint when a participant's finger was bent passively by a motor to extract its natural limits [44]. For the sake of visual clarity, we depicted these values of 0.84 (MCP) as a dashed line in our charts (annotated as "maximum voluntary independence"). Unlike MCP, finger movements around the PIP are more independent from other joints [27], so we abstain from the maximum voluntary independence for PIP to our plots, since these are highly independent of MCP.

6.2 Result#1: dextrEMS doubled finger independence of EMS

Figure 7 (a) depicts our results for all four target fingers, as they were actuated around their MCP and PIP joints **Overall, we found that dextrEMS doubled the finger independence achieved with EMS alone**. We found an average independence of 0.25 (SD = 0.31) when actuating MCP and PIP joints using EMS alone, and an average independence of 0.60 (SD=0.25) when actuating MCP and PIP joints using dextrEMS. This is our main result for this technical evaluation.

Now, we analyze the overall independence for all four fingers around each joint. For the MCP joint of all fingers, we found an average independence of 0.18 (SD=0.17) when actuating MCP joints using EMS alone, and an average independence of 0.56 (SD=0.16) when actuating MCP joints using dextrEMS—as such, on average, for all MCP joints, dextrEMS almost tripled the finger independence of EMS. Secondly, for the PIP joint of all fingers, we found an average independence of 0.31 (SD=0.40) when actuating PIP joints using EMS alone, and an average independence of 0.65 (SD=0.31) when

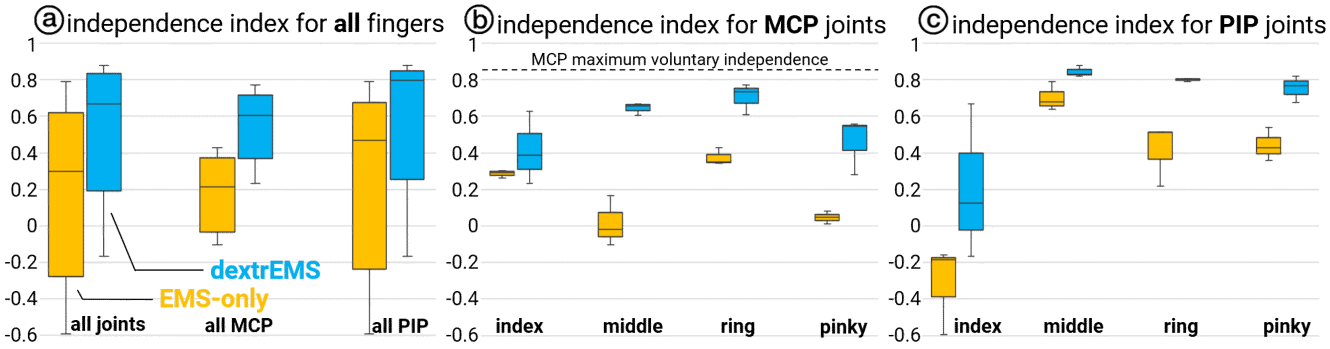

Figure 7: (a) Overall independence for all fingers in both conditions. (b) and (c) Result breakdown per joint (MCP vs. PIP) and finger.

actuating MCP joints using dextrEMS—as such, on average, for all PIP joints, dextrEMS doubled the independence of EMS.

For the MCP joint, we observed improvements in the independence of all fingers, except the index. Then, for the PIP joints, we observed improvements in the independence of all fingers. Remarkably, even the ring finger, known to be "robust to actuate without parasitical motions of adjacent fingers" [35], was improved.

Finally, to give the reader a visual sense of the difference in quality between a pose rendered by dextrEMS when compared to EMS-only, we provide exemplary situations taken directly from the camera data. Figure 8 depicts an exemplary improvement over an independent flexion of the ring finger at the PIP joint; this joint is notoriously hard to address using EMS-only. Indeed, we found that using EMS-only, this finger (even with the best calibration and placement) tends to unwantedly flex its own MCP and the MCP and PIP of the adjacent ring finger.

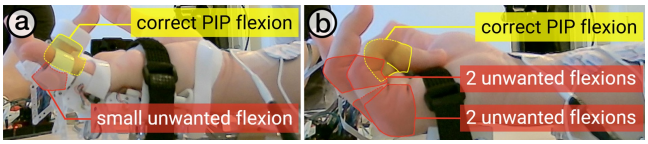

Figure 8: An example of how (a) dextrEMS improved the dexterity of a PIP flexion of the ring finger when compared to (b) EMS-only.

Figure 9 depicts an exemplary improvement over an independent flexion of the index finger at the MCP joint; this is an easier finger to address using EMS-only but not independently of unwanted PIP movements from other fingers. Indeed, we found that using dextrEMS was able to reduce unwanted movements.

6.3 Technical evaluation#2: Pose precision (i.e., stopping the finger at precise angles without oscillations)

In this test, we focus on measuring dextrEMS' ability to stop the finger at a target angle **without oscillations**. In this technical evaluation, we measured the precision and unwanted oscillations caused by dextrEMS and EMS-alone in stopping a finger at 12

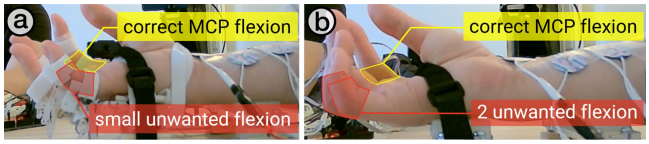

Figure 9: An example of how (a) dextrEMS improved the dexterity of an MCP flexion of the index finger when compared to (b) EMS-only.

different angles. We chose to actuate the ring finger around the MCP joint, since it is easy to actuate (i.e., it is easy to move using EMS, as used in [35]). This finger is also *extremely* sensitive to EMS currents and hard to stop at a precise angle without oscillations; in other words, this finger *is an ideal case* to see if dextrEMS can improve its performance at the MCP joint. Moreover, as our evaluation involved 120 trials for a single finger, evaluating the performance of all remaining four fingers would likely extend the study duration dramatically and induce fatigue in participants.

Interface conditions. As such, we compared: **(1) dextrEMS** device and, (2) the **existing EMS** approach, which is based on a PID controller that regulates the contraction of opposing muscles to halt the finger at a precise angle.; this approach is replicated from [49] and used widely in interactive devices based of EMS [37, 43, 49, 72, 74, 88].

Apparatus. We use the same apparatus from our previous technical evaluation. Additionally, because the haptic devices aimed to stop the finger at a precise angle, they required a closed-loop

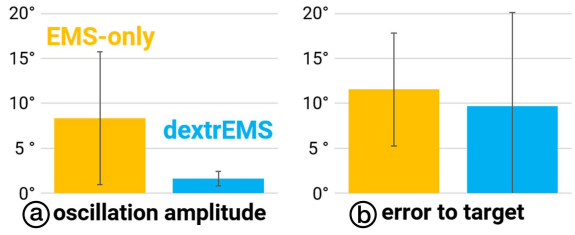

Figure 10: (a) Amplitude of oscillations and (b) error to target in stopping a ring finger with dextrEMS or EMS, for all 12 angles.

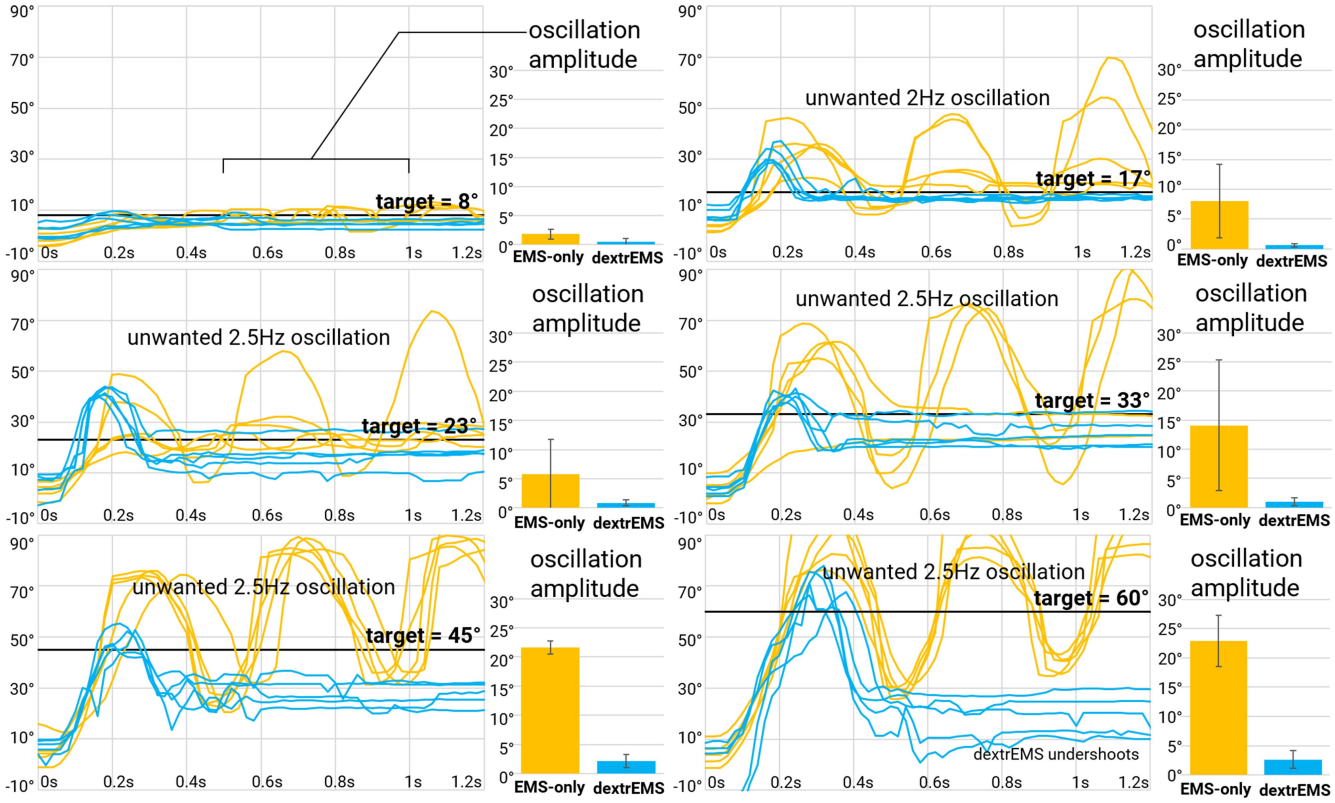

Figure 11: Accuracy (error to target) in stopping a flexing ring finger with dextrEMS or EMS, for six angles. Each plot is annotated with the frequency of any observed large oscillations (typically 2-2.5Hz with EMS-only condition), and a sidebar chart with the amplitude of oscillations between 0.5-1s.

(between the current finger angle vs. target angle to reach). As such, we supply both EMS and dextrEMS with the finger pose (angle of MCP) in real-time. To realize this, we used the Optitrack motion capture system to retrieve the angle with two rigid bodies, attached to the proximal phalanx of the target finger and a rigid-body baseline attached at the back of the hand. Note that, prior to this, we compared between this Optitrack motion tracking system and placing two of the aforementioned 9DOF IMUs combined with Madgwick filters [58] and found that the Optitrack yielded a higher accuracy and no drift over time.

Calibration procedure. For each angle measured, we calibrated the EMS placement following our previous test. Then, we iteratively calibrated the PID controllers for each condition. We varied the stimulation pulse-width from 100 to 450 μs, while using the same current intensity (7 mA for flexor and 8 mA for extensor) in both conditions. For the existing EMS condition (baseline), its dual-sided PID (i.e., extensor and flexor) was tuned for maximum precision with minimum oscillations, however, these are not always possible to dissipate entirely. As for dextrEMS condition, a single-sided PID was tuned, in the same manner, to reach the target with maximum precision; moreover, we also calibrated the brake's mechanical latency to compensate for its triggering time. While both conditions were calibrated, to the best of the expert experimenter's knowledge, with the same goal to minimize error, as with most studies involving replication of a baseline by an experimenter, there is a small chance of unconscious bias. Yet, we have ample experience in tuning these types of PID controllers for EMS.

Procedure. We recorded a total of 120 trials: 6 angles x 2 starting positions (rested and open hand) x 5 repetitions x 2 interface conditions. In each trial, the EMS actuated the ring finger to one of the target angles for flexion (8°, 17°, 23°, 33°, 45°, 60°) and extension (0°, 8°, 17°, 23°, 33°, 45°). The angles were defined by the resolution of dextrEMS' ratchet and the anatomy of the participant since the finger joint's articulation does not translate linearly to the ratchet's rotation. The order of the interface conditions and, within it, the order of angles was randomized. As a result, each trial depicts the error between the desired angle and the final angle. Note that our device and the baseline exhibit different stopping behaviors: (1) dextrEMS uses the brake to stop the finger at the target angle, when this is locked, the angle thus remains final (with some variance due to mechanical slack at the Velcro connections); and, (2) the canonical EMS approach to stop a moving finger (as well as other limbs) relies on controllers (typically a PID, e.g., [49]) that regulate a dual-sided muscle stimulation, i.e., the controller regulates the intensity of EMS contractions of the flexor and extensor, so that, in theory, these cancel out and stop the finger from moving. In practice, even with our best efforts and years of experience in calibrating PID for EMS, these tunings often still result in some

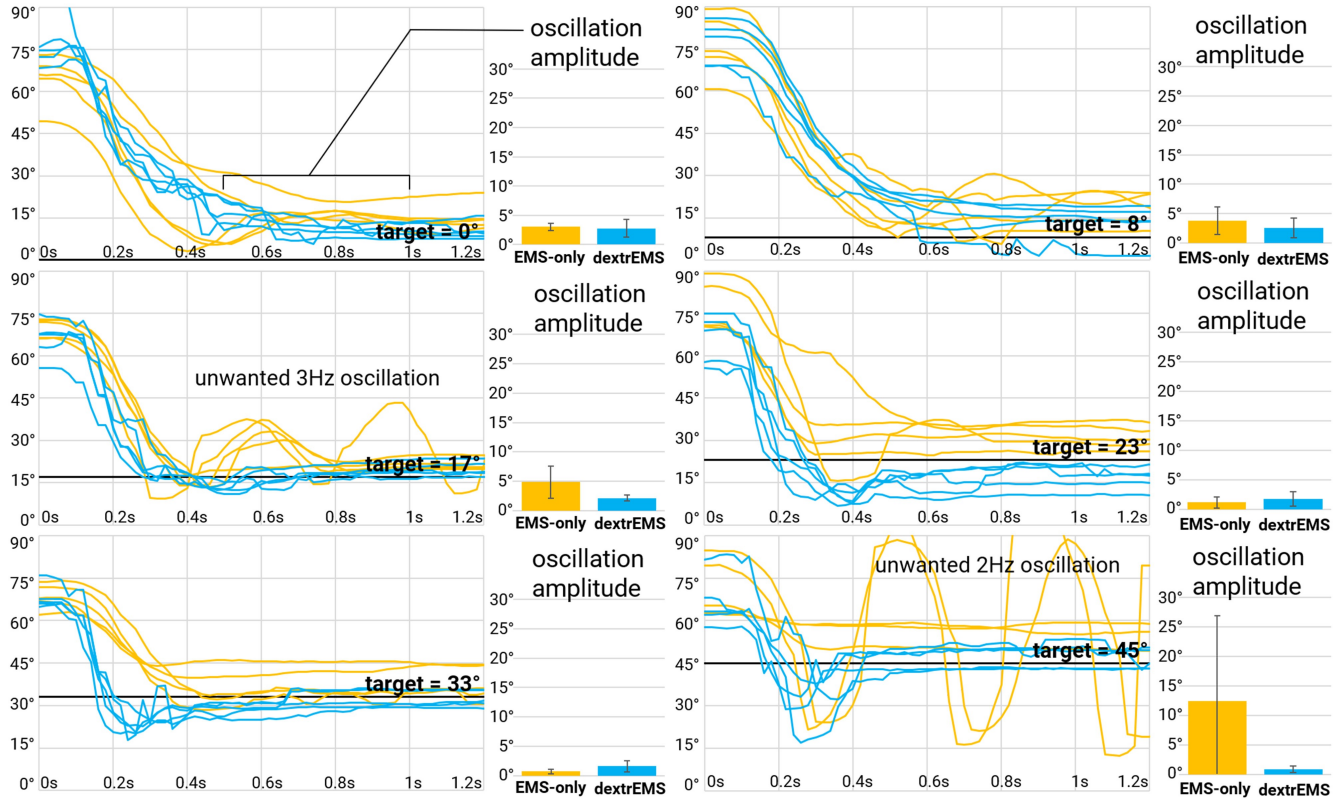

Figure 12: Accuracy (error to target) in stopping an extending ring finger with dextrEMS or EMS, for six angles. Each plot is annotated with the frequency of any observed large oscillations (typically 2-3Hz with EMS-only condition), and a sidebar chart with the amplitude of oscillations between 0.5-1s.

oscillation, and not in an entirely "stopped" finger. To capture this dynamic behavior and reveal more about the two haptic devices, we recorded the angle data for 1.2 seconds.

6.4 Results#2: dextrEMS holds fingers without oscillations experienced with EMS only

We found that for all 12 angles, dextrEMS minimized the oscillations typical of PID controllers used in EMS; this is the main finding of this experiment, which is depicted in Figure 10. Oscillations were measured by analyzing the standard deviation of each trial and averaging these per condition for the time frame between 0.5s-1.0s. We found an average oscillation amplitude of 1.62° (SD=0.81°) when actuated by dextrEMS, and an average oscillation amplitude of 8.37° (SD=7.39°) when actuated by EMS-only.

Then, also in Figure 10, we depict the average error to the target for each condition, all angles. We found an average error of 9.72° (SD=10.37°) when actuated by dextrEMS, and an average error of 11.55° (SD=6.23°) when actuated by EMS-only. As such, we found these to be comparable. What is beneficial to our approach is that while EMS-only oscillates around the target with an average error of 11.55°, dextrEMS brakes around the target with an average error of 9.72°; as such, the improvement is dextrEMS does so *without unwanted oscillations.*

Moreover, as discussed before, dextrEMS average error to target is derived from its ratchet's resolution, which is 24 teeth in our simple laser cut ratchet. One can easily double (or more) the resolution of dextrEMS by CNC-machining the ratchet out of stronger materials (e.g., metal rather than simple acrylic).

In Figure 11, we depict the trajectories of the ring finger, as it flexes around the MCP joint, for all six angles. Again, we found that dextrEMS minimizes unwanted oscillations caused by the EMS' PID controllers. Moreover, it also depicts where dextrEMS excels and where it can still be improved in terms of halting finger flexions at precise angles: dextrEMS was braking in a stable manner for three of six angles (i.e., for 17°, 23°, and 33°) while in the at 45° and 60 ° we observed the dextrEMS brake undershooting on occasion.

Finally, in Figure 12, we depict the trajectories of the ring finger, as it extends around the MCP joint, for all six angles. We found that dextrEMS minimized unwanted oscillations from EMS' PID controllers. Moreover, we depicted where dextrEMS excels in terms of halting finger extensions at precise angles: dextrEMS provided superior accuracy for 17° and 45° targets, which in case the EMS suffered from oscillations. In the remainder angles, we observed the dextrEMS and EMS to have similar performance for extensions, especially at 0° and 8° where both dextrEMS and EMS displayed a similar undershooting profile.

6.5 Technical evaluation#3: Latency

To measure the end-to-end latency of our device, we used a high-speed camera at 240 fps to film a particular joint of our device as well as the screen of the VR application, which was set to flash once a message to our device was sent; this message informed one joint to lock using the brake. We obtained an end-to-end latency of 229.2ms, which we are able to break down into: 176.7ms of WiFi communication latency (from VR to a PC via a WiFi router), 40ms of BLE communication latency (from a PC to our microcontroller receiving this message and blinking a built-in LED in response) and 12.5ms of mechanical latency (from our motor driver LED turning on to the brake being fully locked with the ratchet). While 229.2ms is not appropriate for fast haptic feedback, note that the key latency here is only 52.5ms (from receiving a BLE message to locking a joint), which is suitable for most haptic applications. The remaining 176.7ms of latency from the WiFi communication was not optimized in our applications.

6.6 Technical evaluation#4: Maximum braking force

Finally, we measured the maximum braking force of our mechanical brakes. To do so, we used a load cell rated for 5kg with an ADC amplifier (HX711), which was sampled at 100Hz via a 16-bit microcontroller. The base of the load cell was clamped onto a rigid support structure, while the other end was floating with the exoskeleton secured on it. We tested both joints separately to identify the failure point of the brake with their linkages set to the longest setting. In both tests, we locked the joint's brake and applied an increasing force. We sampled the force sensor until the brake mechanism was broken. We found a maximum braking force of 1377.4g (13.5N) for the MCP joint and 1348.1g (13.2N) for the PIP with our 3D printed resin braking mechanism. Naturally, we expect that printing our design using robust materials such as carbon fiber or aluminum would yield a considerably larger braking force. On the other hand, our 3D printed mechanism already absorbs 13N, which is reasonable for the proposed applications.

Additionally, to assist in understanding the braking force that our mechanism provides, we also measured how much force a finger that is stimulated by EMS exhibited. We found an average force of 11.9N (SD=0.25N) across all four fingers, using three repetitions at an EMS intensity 1mA higher than the maximum used in our evaluations. Furthermore, these measurements were done with both PIP and DIP joints combined, which stands in contrast to our braking force evaluation, which tested for one joint at a time. To sum up, the average forced-induced by EMS on the finger (11.9N) was below our braking force (13.5N), which explained why no dextrEMS broke in our evaluations.

7 APPLICATIONS

We created five applications that showcase our haptic device: (1) haptic fingerspelling of a small subset of ASL letters; haptic guidance for guitar chords; (3) VR piano; (4) whack-a-mole game; and, (5) VR bouldering simulator.

7.1 Application #1: haptic fingerspelling for a small subset of American Sign Language (ASL)

Figure 13 depicts a mobile application that enables a user, without prior knowledge of fingerspelling the manual ASL alphabet, to use dextrEMS to communicate simple letters to a hard-of-hearing individual. As aforementioned, dextrEMS does not solve the inherent challenges faced by the hard of hearing individual or deaf community, not only because of its limited abilities but, also, because no technological solution should "solve" the unique experience of these individuals. Instead, we were inspired by the ASL alphabet and use it to demonstrate dextrEMS.

Figure 13 (a) shows the non-ASL cognizant user pointing their smartphone's camera to their friend, who is speaking using letters of a fingerspelling alphabet. The smartphone is running our custom application to illustrate the portability of dextrEMS. Our application leverages *MediaPipe* hand tracking [56] to recognize a fixed set of pre-trained and simple ASL fingerspelled signs. Once a letter sign is recognized, it is shown on the user's smartphone as text, depicting that their friend said, "what are you studying?" in ASL. Figure 13 (b), shows how the user, who unfortunately does not know how to fingerspell, types "HCI" as a response using the phone's keyboard. Our application then translates the text, according to ASL fingerspelling, into hardware instructions for dextrEMS. The instruction is then sent to dextrEMS via Bluetooth. In response, dextrEMS renders each letter's corresponding pose by actuating the

Figure 13: Our custom smartphone application detects fingerspelling by using MediaPipe and translates it into text. (b) The user types its response "HCI" onto the smartphone, which (b, c, and d) dextrEMS outputs by actuating the fingers to form a fingerspelled "HCI".

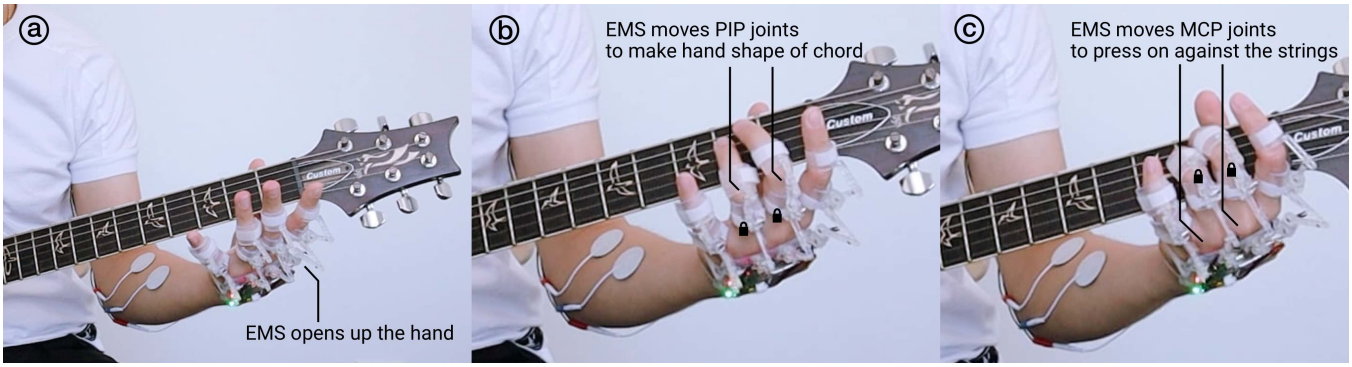

Figure 14: In this simple application, dextrEMS assists a user in performing the E-minor chord, which we chose due to its pose simplicity.

user's muscles with sufficient dexterity to convey these, relatively simple, letters. Figure 13 (b) shows dextrEMS outputting the "H" of "HCI". After rendering the "H" letter for three seconds, dextrEMS moves to the next letter ("C" in "HCI"), which is depicted in Figure 13 (c). To return the user's hand to its resting state it releases all the locks are released and dextrEMS makes use of all the extensors to "reset" the hand to its resting pose. Finally, as shown in Figure 13 (d), dextrEMS repeats the same process to fingerspell the "I" in "HCI". Again, we highlight that both the complete set of ASL fingerspelled letters is of an immense complexity that neither dextrEMS nor any device can achieve, but also sign language entails a vast richness that simple devices like ours will never have the capacity to assist with (including hand gestures, body language, facial expressions and more).

7.2 Application #2: guitar chords using haptic guidance

Our second application explores a different haptic tutorial: guitar chords. Guitar requires great dexterity, especially when novice players explore where to place their fingers on the fretboard to form chords. In this application, depicted in Figure 14, we demonstrate how dextrEMS assists a player in forming a chord by actuating their fingers to the correct position on the fretboard to form an E minor chord. Note that while this chord is relatively simple (e.g., compared to a more complex chord, such as an E minor flat sixth), it would not be easily achievable with EMS alone.

To position the fingers at the chord, dextrEMS first extends all fingers (brakes open) so that none are touching the fretboard. Then, it locks the fingers that will not be used in the E-minor chord. Before actuating the MCPs, it adjusts the fingers' PIP joints' angle using EMS actuation and braking when the PIP joints are in place. Once the chord finger has been formed in mid-air, dextrEMS unlocks the MCPs and pushes all fingers with EMS against the fretboard.

7.3 Application #3: a force feedback piano chord tutor in VR

In Figure 15, we demonstrate how dextrEMS enables our VR piano application to independently actuate each finger to achieve simple melodies or even multiple notes. First, dextrEMS actuates the user's fingers by locking those that should not move and actuating the remainder via EMS to form the target notes; here, it actuates the user's hand to play a minor third interval over D, which is depicted in Figure 15 (b). When the user's fingers reach the bottom of the piano keys (i.e., the virtual keys' mechanism is bottomed out and cannot move more), dextrEMS locks all fingers, providing the haptic feedback of the final position, which is depicted in Figure 15 (c).

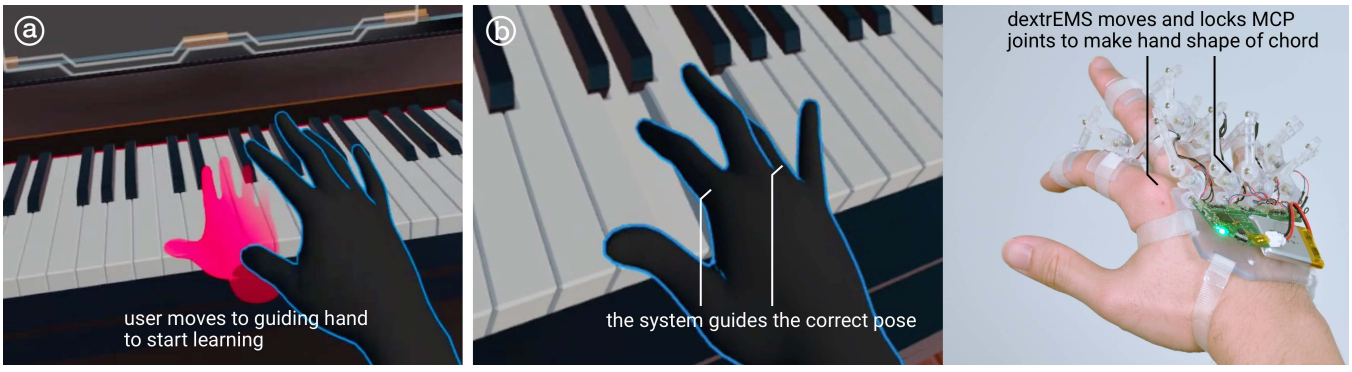

Figure 15: In this simple application, dextrEMS assists a user in performing a minor third interval of D.

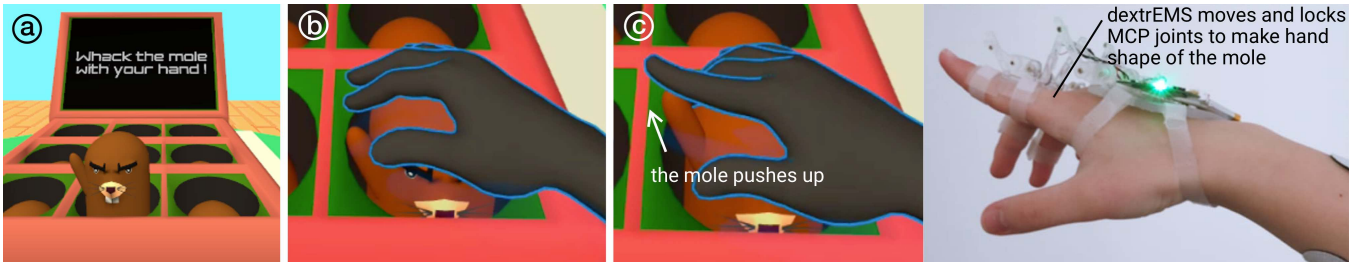

Figure 16: (a) In our whack-a-mole game, (b) players can feel the shape of the mole; (c) here, the mole fights back by extending its arm upwards, which dextrEMS renders by actuating the index and locking it at the correct pose.

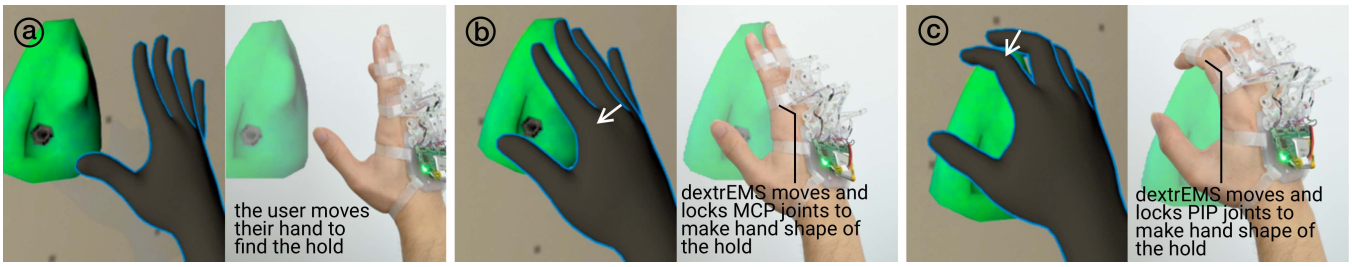

Figure 17: (a) This user swipes their hand sideways across the bouldering wall, trying to feel the next hold. (b) as their hand reaches over the hold, it starts to be actuated by dextrEMS to pose the hand to (c) emulate the different hold shapes.

7.4 Application #4: feeling the *changing shape* of the moles in a VR whack-a-mole game

We provide an example of a more traditional use of EMS as force-feedback in VR. However, instead of using EMS for large and coarse movements (as typical of EMS in VR [48, 52]), we use dextrEMS for rendering a changing "shape", which requires more dexterous finger actuation. To demonstrate this, we implemented a simple *Whack-A-Mole* game in VR, consisting of pushing the moles back to their holes by whacking them on their heads. While in the typical rendition of this game moles disappear once hit, our moles fight back by pushing one of their arms up, pushing the user's index finger up. We make use of dextrEMS to render this individual motion of the finger (dextrEMS locks all other joints and actuates the extensors, causing only the index finger to move up). As the mole fully extends their arm, dextrEMS now locks the index as well, resulting in a final pose that matches the expected shape of the mole. The user holds the mole for a few seconds, and it recoils to the hole. Now, dextrEMS releases all brakes, and the user's finger falls back down. Finally, the user repeats this for the remaining moles, to win the game.

7.5 Application #5: VR bouldering simulator

In this last VR experience, we demonstrate how dextrEMS allows this user to feel the shapes of bouldering holds (the protuberances that stand out from a bouldering wall). Figure 17 (a) shows the user moving their hand across the wall, trying to find the next hold to grab onto. As their fingers first hit the side of the hold, dextrEMS actuates their index finger upwards and locks it into the height of the shape, which is depicted in Figure 17 (b). Then, dextrEMS continues to actuate and lock remainder fingers, until they are in the correct geometry of the bouldering hold. Figure 17 (c) shows the user's fingers taking the shape of the hold. Conversely, as the user continues moving to the other side, the process reverses and dextrEMS actuates the fingers down to the wall.

8 CONCLUSION

In this paper, we increased the dexterity of electrical muscle stimulation, by proposing and engineering an EMS-based haptic device combined with a lightweight mechanical brake attached to each finger joint, which we called dextrEMS. The key idea behind dextrEMS is that while the EMS actuates the fingers, it is our mechanical brake that *stops* the finger in the correct pose. Moreover, dextrEMS also uses its brakes to select which fingers are "free" to be moved by means of EMS, eliminating unwanted movements by preventing adjacent fingers from moving.

We implemented dextrEMS as an untethered haptic device that actuates eight finger joints independently (metacarpophalangeal and proximal interphalangeal joints for four fingers), which we demonstrate in a wide range of haptic applications, such as assisted fingerspelling, a piano tutorial, guitar tutorial, and a VR game.

Finally, in our technical evaluation, we found that dextrEMS outperformed EMS alone by doubling its independence and reducing unwanted oscillations.

As for future work, we expect researchers following up on dextrEMS to create variations attached to other body joints; in other words, dextrEMS is ripe for applications beyond just finger actuation, for instance, one can attach a brake to the elbow joint and precisely brake the biceps without the need for EMS-PID controllers for the triceps (as in [49, 52]). Moreover, there are several ways researchers can expand on our mechanical design. For instance,

researchers might explore replacing our ratchet-pawl mechanism with a truly bidirectional ratchet, which involves more logic and mechanics (thus, potentially larger), but will enable even quicker stopping in the extensor direction. Similarly, we expect other researchers to fabricate dextrEMS using even stronger material (e.g., CNC machined ratchets made from metal), which easily leads to doubling its resolution, which is primarily only limited by the resolution of the ratchet teeth. Finally, other researchers might explore adding our mechanism to the DIP joint, to enable new applications that use even finer motor skills of all three finger joints.

ACKNOWLEDGMENTS

This project initially started at McGill University under Prof. Cooperstock's supervision in collaboration with Romain Nith, Clement Fournier and Augustin Legrand. We would like to thank Prof. Cooperstock for his valuable feedback, and the rest of the team for their contribution and support.

REFERENCES

[1] Bao, X., Zhou, Y., Wang, Y., Zhang, J., Lü, X. and Wang, Z. 2018. Electrode placement on the forearm for selective stimulation of finger extension/flexion. *PLOS ONE*. 13, 1 (Jan. 2018), e0190936. DOI:https://doi.org/10.1371/journal.pone.0190936.

[2] BeagleBoard.org - pocket: https://beagleboard.org/pocket. Accessed: 2021-04-07.

[3] Bergamasco, M., Allotta, B., Bosio, L., Ferretti, L., Parrini, G., Prisco, G.M., Salsedo, F. and Sartini, G. 1994. An arm exoskeleton system for teleoperation and virtual environments applications. *Proceedings of the 1994 IEEE International Conference on Robotics and Automation* (May 1994), 1449–1454 vol.2.

[4] Blake, J. and Gurocak, H.B. 2009. Haptic Glove With MR Brakes for Virtual Reality. *IEEE/ASME Transactions on Mechatronics*. 14, 5 (Oct. 2009), 606–615. DOI:https://doi.org/10.1109/TMECH.2008.2010934.

[5] Bouzit, M., Burdea, G., Popescu, G. and Boian, R. 2002. The Rutgers Master II-new design force-feedback glove. *IEEE/ASME Transactions on Mechatronics*. 7, 2 (Jun. 2002), 256–263. DOI:https://doi.org/10.1109/TMECH.2002.1011262.

[6] Carignan, C., Tang, J. and Roderick, S. 2009. Development of an exoskeleton haptic interface for virtual task training. *2009 IEEE/RSJ International Conference on Intelligent Robots and Systems* (Oct. 2009), 3697–3702.

[7] Chen, K., Zhang, B. and Zhang, D. 2014. Master-Slave Gesture Learning System Based on Functional Electrical Stimulation. *Intelligent Robotics and Applications*. X. Zhang, H. Liu, Z. Chen, and N. Wang, eds. Springer International Publishing. 214–223.

[8] Choi, I., Corson, N., Peiros, L., Hawkes, E.W., Keller, S. and Follmer, S. 2018. A Soft, Controllable, High Force Density Linear Brake Utilizing Layer Jamming. *IEEE Robotics and Automation Letters*. 3, 1 (Jan. 2018), 450–457. DOI:https://doi.org/10.1109/LRA.2017.2761938.

[9] Choi, I., Culbertson, H., Miller, M.R., Olwal, A. and Follmer, S. 2017. Grabity: A Wearable Haptic Interface for Simulating Weight and Grasping in Virtual Reality. *Proceedings of the 30th Annual ACM Symposium on User Interface Software and Technology* (New York, NY, USA, 2017), 119–130.

[10] Choi, I. and Follmer, S. 2016. Wolverine: A Wearable Haptic Interface for Grasping in VR. *Proceedings of the 29th Annual Symposium on User Interface Software and Technology* (New York, NY, USA, 2016), 117–119.

[11] Colley, A., Leinonen, A., Forsman, M.-T. and Häkkilä, J. 2018. EMS Painter: Co-creating Visual Art using Electrical Muscle Stimulation. *Proceedings of the Twelfth International Conference on Tangible, Embedded, and Embodied Interaction* (New York, NY, USA, Mar. 2018), 266–270.

[12] CyberGrasp: http://www.cyberglovesystems.com/cybergrasp. Accessed: 2019-09-02.

[13] Dingler, T., Goto, T., Tag, B. and Kunze, K. 2017. EMS icons: conveying information by analogy to enhance communication through electrical muscle stimulation. *Proceedings of the 2017 ACM International Joint Conference on Pervasive and Ubiquitous Computing and Proceedings of the 2017 ACM International Symposium on Wearable Computers* (New York, NY, USA, Sep. 2017), 732–739.

[14] Duente, T., Pfeiffer, M. and Rohs, M. 2016. On-skin technologies for muscle sensing and actuation. *Proceedings of the 2016 ACM International Joint Conference on Pervasive and Ubiquitous Computing: Adjunct* (New York, NY, USA, Sep. 2016), 933–936.

[15] Duente, T., Pfeiffer, M. and Rohs, M. 2017. Zap++: a 20-channel electrical muscle stimulation system for fine-grained wearable force feedback. *Proceedings of the 19th International Conference on Human-Computer Interaction with Mobile Devices and Services* (New York, NY, USA, Sep. 2017), 1–13.

[16] Duente, T., Schneegass, S. and Pfeiffer, M. 2017. EMS in HCI: challenges and opportunities in actuating human bodies. *Proceedings of the 19th International Conference on Human-Computer Interaction with Mobile Devices and Services* (New York, NY, USA, Sep. 2017), 1–4.

[17] Duente, T., Schulte, J., Pfeiffer, M. and Rohs, M. 2018. MuscleIO: Muscle-Based Input and Output for Casual Notifications. *Proceedings of the ACM on Interactive, Mobile, Wearable and Ubiquitous Technologies*. 2, 2 (Jul. 2018), 64:1-64:21. DOI:https://doi.org/10.1145/3214267.

[18] Ebisu, A., Hashizume, S. and Ochiai, Y. 2018. Building a feedback loop between electrical stimulation and percussion learning. *ACM SIGGRAPH 2018 Studio* (Vancouver, British Columbia, Canada, Aug. 2018), 1–2.

[19] Ebisu, A., Hashizume, S., Suzuki, K., Ishii, A., Sakashita, M. and Ochiai, Y. 2017. Stimulated percussions: method to control human for learning music by using electrical muscle stimulation. *Proceedings of the 8th Augmented Human International Conference on - AH '17* (Silicon Valley, California, 2017), 1–5.

[20] ExoHand | Festo Corporate: https://www.festo.com/group/en/cms/10233.htm. Accessed: 2021-04-06.

[21] Farbiz, F., Yu, Z.H., Manders, C. and Ahmad, W. 2007. An Electrical Muscle Stimulation Haptic Feedback for Mixed Reality Tennis Game. *ACM SIGGRAPH 2007 Posters* (New York, NY, USA, 2007).

[22] Fontana, M., Ruffaldi, E., Salasedo, F. and Bergamasco, M. On the Integration of Tactile and Force Feedback. *Haptics Rendering and Applications*.

[23] Fortin, P.E., Blum, J.R. and Cooperstock, J.R. 2017. Raising the Heat: Electrical Muscle Stimulation for Simulated Heat Withdrawal Response. *Adjunct Publication of the 30th Annual ACM Symposium on User Interface Software and Technology* (New York, NY, USA, Oct. 2017), 137–139.

[24] Grönvall, E., Fritsch, J. and Vallgårda, A. 2016. FeltRadio: Sensing and Making Sense of Wireless Traffic. *Proceedings of the 2016 ACM Conference on Designing Interactive Systems - DIS '16* (Brisbane, QLD, Australia, 2016), 829–840.

[25] Gu, X., Zhang, Y., Sun, W., Bian, Y., Zhou, D. and Kristensson, P.O. 2016. Dexmo: An Inexpensive and Lightweight Mechanical Exoskeleton for Motion Capture and Force Feedback in VR. *Proceedings of the 2016 CHI Conference on Human Factors in Computing Systems* (New York, NY, USA, 2016), 1991–1995.

[26] Gui, K. and Zhang, D. 2014. Influence of volitional contraction on muscle response to functional electrical stimulation. *2014 IEEE 19th International Functional Electrical Stimulation Society Annual Conference (IFESS)* (Sep. 2014), 1–4.

[27] Häger-Ross, C. and Schieber, M.H. 2000. Quantifying the Independence of Human Finger Movements: Comparisons of Digits, Hands, and Movement Frequencies. *Journal of Neuroscience*. 20, 22 (Nov. 2000), 8542–8550. DOI:https://doi.org/10.1523/JNEUROSCI.20-22-08542.2000.

[28] Hanagata, S. and Kakehi, Y. 2018. Paralogue: A Remote Conversation System Using a Hand Avatar which Postures are Controlled with Electrical Muscle Stimulation. *Proceedings of the 9th Augmented Human International Conference* (New York, NY, USA, Feb. 2018), 1–3.

[29] HaptX | Haptic gloves for VR training, simulation, and design: https://haptx.com/. Accessed: 2020-05-04.

[30] Hassib, M., Pfeiffer, M., Schneegass, S., Rohs, M. and Alt, F. 2017. Emotion Actuator: Embodied Emotional Feedback Through Electroencephalography and Electrical Muscle Stimulation. *Proceedings of the 2017 CHI Conference on Human Factors in Computing Systems* (New York, NY, USA, 2017), 6133–6146.

[31] Hinchet, R., Vechev, V., Shea, H. and Hilliges, O. 2018. DextrES: Wearable Haptic Feedback for Grasping in VR via a Thin Form-Factor Electrostatic Brake. *Proceedings of the 31st Annual ACM Symposium on User Interface Software and Technology* (New York, NY, USA, 2018), 901–912.

[32] Hummelsheim, H., Maier-Loth, M.L. and Eickhof, C. 1997. The functional value of electrical muscle stimulation for the rehabilitation of the hand in stroke patients. *Scandinavian Journal of Rehabilitation Medicine*. 29, 1 (Mar. 1997), 3–10.

[33] Ishimaru, T. and Saga, S. 2020. Virtual bumps display based on electrical muscle stimulation. *2020 IEEE Haptics Symposium (HAPTICS)* (Mar. 2020), 96–101.

[34] Jain, S., Sharma, S. and Babbar, D. 2017. Star-Force: A Playful Implementation of the Jedi-force. *Proceedings of the Tenth International Conference on Tangible, Embedded, and Embodied Interaction - TEI '17* (Yokohama, Japan, 2017), 761–766.

[35] Kasahara, S., Nishida, J. and Lopes, P. 2019. Preemptive Action: Accelerating Human Reaction Using Electrical Muscle Stimulation Without Compromising Agency. *Proceedings of the 2019 CHI Conference on Human Factors in Computing Systems* (New York, NY, USA, 2019), 643:1-643:15.

[36] Katoh, M., Nishimura, N., Yokoyama, M., Hachisu, T., Sato, M., Fukushima, S. and Kajimoto, H. 2013. Optimal selection of electrodes for muscle electrical stimulation using twitching motion measurement. *Proceedings of the 4th Augmented Human International Conference on - AH '13* (Stuttgart, Germany, 2013), 237–238.

[37] Kaul, O.B., Pfeiffer, M. and Rohs, M. 2016. Follow the Force: Steering the Index Finger towards Targets using EMS. *Proceedings of the 2016 CHI Conference Extended Abstracts on Human Factors in Computing Systems* (New York, NY, USA, May 2016), 2526–2532.

[38] Kitamura, T., Mizoguchi, H., Mizukami, N., Sakaino, S. and Tsuji, T. 2017. Chattering reduction of functional electrical stimulation with the smith compensator. *IECON 2017 - 43rd Annual Conference of the IEEE Industrial Electronics Society* (Oct. 2017), 7577–7582.

[39] Knibbe, J., Strohmeier, P., Boring, S. and Hornbæk, K. 2017. Automatic Calibration of High Density Electric Muscle Stimulation. *Proceedings of the ACM on Interactive, Mobile, Wearable and Ubiquitous Technologies*. 1, 3 (Sep. 2017), 1–17. DOI:https://doi.org/10.1145/3130933.

[40] Kono, M., Ishiguro, Y., Miyaki, T. and Rekimoto, J. 2018. Design and Study of a Multi-Channel Electrical Muscle Stimulation Toolkit for Human Augmentation. *Proceedings of the 9th Augmented Human International Conference* (New York, NY, USA, Feb. 2018), 1–8.

[41] Kono, M., Takahashi, T., Nakamura, H., Miyaki, T. and Rekimoto, J. 2018. Design Guideline for Developing Safe Systems that Apply Electricity to the Human Body. *ACM Transactions on Computer-Human Interaction*. 25, 3 (Jun. 2018), 19:1-19:36. DOI:https://doi.org/10.1145/3184743.

[42] Kruijff, E., Schmalstieg, D. and Beckhaus, S. 2006. Using neuromuscular electrical stimulation for pseudo-haptic feedback. *Proceedings of the ACM symposium on Virtual reality software and technology* (New York, NY, USA, Nov. 2006), 316–319.

[43] Kurita, Y., Ishikawa, T. and Tsuji, T. 2016. Stiffness Display by Muscle Contraction Via Electric Muscle Stimulation. *IEEE Robotics and Automation Letters*. 1, 2 (Jul. 2016), 1014–1019. DOI:https://doi.org/10.1109/LRA.2016.2529689.

[44] Lang, C.E. and Schieber, M.H. 2004. Human Finger Independence: Limitations due to Passive Mechanical Coupling Versus Active Neuromuscular Control. *Journal of Neurophysiology*. 92, 5 (Nov. 2004), 2802–2810. DOI:https://doi.org/10.1152/jn.00480.2004.

[45] Limanowski, J., Lopes, P., Keck, J., Baudisch, P., Friston, K. and Blankenburg, F. 2020. Action-Dependent Processing of Touch in the Human Parietal Operculum and Posterior Insula. *Cerebral Cortex*. 30, 2 (Mar. 2020), 607–617. DOI:https://doi.org/10.1093/cercor/bhz111.

[46] Lopes, P. and Baudisch, P. 2017. Immense Power in a Tiny Package: Wearables Based on Electrical Muscle Stimulation. *IEEE Pervasive Computing*. 16, 3 (2017), 12–16. DOI:https://doi.org/10.1109/MPRV.2017.2940953.

[47] Lopes, P. and Baudisch, P. 2013. Muscle-propelled Force Feedback: Bringing Force Feedback to Mobile Devices. *Proceedings of the SIGCHI Conference on Human Factors in Computing Systems* (New York, NY, USA, 2013), 2577–2580.

[48] Lopes, P., Ion, A. and Baudisch, P. 2015. Impacto: Simulating Physical Impact by Combining Tactile Stimulation with Electrical Muscle Stimulation. *Proceedings of the 28th Annual ACM Symposium on User Interface Software & Technology* (New York, NY, USA, 2015), 11–19.

[49] Lopes, P., Ion, A., Mueller, W., Hoffmann, D., Jonell, P. and Baudisch, P. 2015. Proprioceptive Interaction. *Proceedings of the 33rd Annual ACM Conference on Human Factors in Computing Systems* (New York, NY, USA, 2015), 939–948.

[50] Lopes, P., Jonell, P. and Baudisch, P. 2015. Affordance++: Allowing Objects to Communicate Dynamic Use. *Proceedings of the 33rd Annual ACM Conference on Human Factors in Computing Systems* (New York, NY, USA, 2015), 2515–2524.

[51] Lopes, P. and Tang, A. 2019. *Rehamove Integration Lib.*

[52] Lopes, P., You, S., Cheng, L.-P., Marwecki, S. and Baudisch, P. 2017. Providing Haptics to Walls & Heavy Objects in Virtual Reality by Means of Electrical Muscle Stimulation. *Proceedings of the 2017 CHI Conference on Human Factors in Computing Systems* (New York, NY, USA, 2017), 1471–1482.

[53] Lopes, P., You, S., Ion, A. and Baudisch, P. 2018. Adding Force Feedback to Mixed Reality Experiences and Games Using Electrical Muscle Stimulation. *Proceedings of the 2018 CHI Conference on Human Factors in Computing Systems* (New York, NY, USA, 2018), 446:1-446:13.

[54] Lopes, P., Yüksel, D., Guimbretière, F. and Baudisch, P. 2016. Muscle-plotter: An Interactive System based on Electrical Muscle Stimulation that Produces Spatial Output. (2016), 207–217.

[55] Lou, Z., Yao, P. and Zhang, D. 2012. Wireless Master-Slave FES Rehabilitation System Using sEMG Control. *Intelligent Robotics and Applications* (Berlin, Heidelberg, 2012), 1–10.

[56] Lugaresi, C., Tang, J., Nash, H., McClanahan, C., Uboweja, E., Hays, M., Zhang, F., Chang, C.-L., Yong, M.G., Lee, J., Chang, W.-T., Hua, W., Georg, M. and Grundmann, M. 2019. MediaPipe: A Framework for Building Perception Pipelines. *arXiv:1906.08172 [cs]*. (Jun. 2019).

[57] MA, Z. and Ben-Tzvi, P. 2015. RML Glove—An Exoskeleton Glove Mechanism With Haptics Feedback. *IEEE/ASME Transactions on Mechatronics*. 20, 2 (Apr. 2015), 641–652. DOI:https://doi.org/10.1109/TMECH.2014.2305842.

[58] Madgwick, S.O.H. An efficient orientation filter for inertial and inertial/magnetic sensor arrays. 32.

[59] Massie, T.H. and Salisbury, J.K. The PHANToM Haptic Interface: A Device for Probing Virtual Objects. 7.

[60] Murayama, J., Bougrila, L., Luo, Y., Akahane, K., Hasegawa, S., Hirsbrunner, B. and Sato, M. SPIDAR G&G: A Two-Handed Haptic Interface for Bimanual VR Interaction. 9.

[61] Nagashima, Y. 2003. Bio-sensing systems and bio-feedback systems for interactive media arts. *Proceedings of the 2003 conference on New interfaces for musical expression* (SGP, May 2003), 48–53.

[62] Nishida, J., Kasahara, S. and Suzuki, K. 2017. Wired muscle: generating faster kinesthetic reaction by inter-personally connecting muscles. *ACM SIGGRAPH 2017 Emerging Technologies* (New York, NY, USA, Jul. 2017), 1–2.

[63] Nishida, J. and Suzuki, K. 2017. bioSync: A Paired Wearable Device for Blending Kinesthetic Experience. *Proceedings of the 2017 CHI Conference on Human Factors in Computing Systems* (New York, NY, USA, May 2017), 3316–3327.

[64] Nishida, J., Takahashi, K. and Suzuki, K. 2015. A wearable stimulation device for sharing and augmenting kinesthetic feedback. *Proceedings of the 6th Augmented Human International Conference* (New York, NY, USA, Mar. 2015), 211–212.

[65] Pfeiffer, M., Duente, T. and Rohs, M. 2016. A Wearable Force Feedback Toolkit with Electrical Muscle Stimulation. *Proceedings of the 2016 CHI Conference Extended Abstracts on Human Factors in Computing Systems - CHI EA '16* (San Jose, California, USA, 2016), 3758–3761.

[66] Pfeiffer, M., Dünte, T., Schneegass, S., Alt, F. and Rohs, M. 2015. Cruise Control for Pedestrians: Controlling Walking Direction Using Electrical Muscle Stimulation. *Proceedings of the 33rd Annual ACM Conference on Human Factors in Computing Systems* (New York, NY, USA, 2015), 2505–2514.

[67] Pfeiffer, M., Schneegaß, S. and Alt, F. 2013. Supporting interaction in public space with electrical muscle stimulation. *Proceedings of the 2013 ACM conference on Pervasive and ubiquitous computing adjunct publication* (New York, NY, USA, Sep. 2013), 5–8.

[68] Pfeiffer, M., Schneegass, S., Alt, F. and Rohs, M. 2014. Let me grab this: a comparison of EMS and vibration for haptic feedback in free-hand interaction. *Proceedings of the 5th Augmented Human International Conference* (New York, NY, USA, Mar. 2014), 1–8.

[69] Pfeiffer, M. and Stuerzlinger, W. 2015. 3D Virtual Hand Selection with EMS and Vibration Feedback. *Proceedings of the 33rd Annual ACM Conference Extended Abstracts on Human Factors in Computing Systems* (New York, NY, USA, Apr. 2015), 1361–1366.

[70] Pohl, H., Hornbæk, K. and Knibbe, J. 2018. Wanding Through Space: Interactive Calibration for Electric Muscle Stimulation. *Proceedings of the 9th Augmented Human International Conference on - AH '18* (Seoul, Republic of Korea, 2018), 1–5.

[71] Popović-Bijelić, A., Bijelić, G., Jorgovanović, N., Bojanić, D., Popović, M.B. and Popović, D.B. 2005. Multi-field surface electrode for selective electrical stimulation. *Artificial Organs*. 29, 6 (Jun. 2005), 448–452. DOI:https://doi.org/10.1111/j.1525-1594.2005.29075.x.

[72] Qiu, S., He, F., Tang, J., Xu, J., Zhang, L., Zhao, X., Qi, H., Zhou, P., Cheng, X., Wan, B. and Ming, D. 2014. Intelligent algorithm tuning PID method of function electrical stimulation using knee joint angle. *2014 36th Annual International Conference of the IEEE Engineering in Medicine and Biology Society* (Aug. 2014), 2561–2564.

[73] Rosen, J., Milutinović, D., Miller, L.M., Simkins, M., Kim, H. and Li, Z. 2014. Unilateral and Bilateral Rehabilitation of the Upper Limb Following Stroke via an Exoskeleton. *Neuro-Robotics*. P. Artemiadis, ed. Springer Netherlands. 405–446.

[74] Rouhani, H., Same, M., Masani, K., Li, Y.Q. and Popovic, M.R. 2017. PID Controller Design for FES Applied to Ankle Muscles in Neuroprosthesis for Standing Balance. *Frontiers in Neuroscience*. 11, (2017). DOI:https://doi.org/10.3389/fnins.2017.00347.

[75] Sandoval-Gonzalez, O., Jacinto-Villegas, J., Herrera-Aguilar, I., Portillo-Rodiguez, O., Tripicchio, P., Hernandez-Ramos, M., Flores-Cuautle, A. and Avizzano, C. 2016. Design and Development of a Hand Exoskeleton Robot for Active and Passive Rehabilitation. *International Journal of Advanced Robotic Systems*. 13, 2 (Mar. 2016), 66. DOI:https://doi.org/10.5772/62404.

[76] Sato, K., Minamizawa, K., Kawakami, N. and Tachi, S. 2007. Haptic Telexistence. *ACM SIGGRAPH 2007 Emerging Technologies* (New York, NY, USA, 2007).

[77] Schneegass, S. and Rzayev, R. 2016. Embodied notifications: implicit notifications through electrical muscle stimulation. *Proceedings of the 18th International Conference on Human-Computer Interaction with Mobile Devices and Services Adjunct* (New York, NY, USA, Sep. 2016), 954–959.

[78] Schneegass, S., Schmidt, A. and Pfeiffer, M. 2016. Creating user interfaces with electrical muscle stimulation. *Interactions*. 24, 1 (Dec. 2016), 74–77. DOI:https://doi.org/10.1145/3019606.

[79] Shao, T., Li, X., Yokoi, H. and Zhang, D. 2016. FESleeve: A Functional Electrical Stimulation System with Multi-electrode Array for Finger Motion Control. *Intelligent Robotics and Applications* (Cham, 2016), 191–199.

[80] Strojnik, P., Kralj, A. and Ursic, I. 1979. Programmed Six-Channel Electrical Stimulator for Complex Stimulation of Leg Muscles During Walking. *IEEE Transactions on Biomedical Engineering*. BME-26, 2 (Feb. 1979), 112–116. DOI:https://doi.org/10.1109/TBME.1979.326520.

[81] Surakijboworn, M. and Wannasuphoprasit, W. 2015. Design of a novel finger exoskeleton with a sliding six-bar joint mechanism. *Proceedings of the 6th Augmented Human International Conference on - AH '15* (Singapore, Singapore, 2015), 77–80.

[82] Takahashi, A., Brooks, J., Kajimoto, H. and Lopes, P. Increasing Electrical Muscle Stimulation's Dexterity by means of Back of the Hand Actuation. *Proceedings of the 2021 CHI Conference on Human Factors in Computing Systems*.

[83] Tamaki, E., Chan, T. and Iwasaki, K. 2016. UnlimitedHand: Input and Output Hand Gestures with Less Calibration Time. *Proceedings of the 29th Annual Symposium on User Interface Software and Technology* (New York, NY, USA, Oct. 2016), 163–165.

[84] Tamaki, E., Miyaki, T. and Rekimoto, J. 2011. PossessedHand: techniques for controlling human hands using electrical muscles stimuli. *Proceedings of the SIGCHI Conference on Human Factors in Computing Systems* (2011), 543–552.

[85] Tatsuno, S., Hayakawa, T. and Ishikawa, M. 2016. Comparison of reaction times in response to electrical and Visual Stimulation using a high-speed camera. *2016 IEEE International Conference on Systems, Man, and Cybernetics (SMC)* (Oct. 2016), 001251–001256.

[86] Tatsuno, S., Hayakawa, T. and Ishikawa, M. 2017. Supportive training system for sports skill acquisition based on electrical stimulation. *2017 IEEE World Haptics Conference (WHC)* (Jun. 2017), 466–471.

[87] Tatsuno, S., Hayakawa, T. and Ishikawa, M. 2017. Trajectory adjustment system for learning based on electrical stimulation. *Proceedings of the 8th Augmented Human International Conference* (New York, NY, USA, Mar. 2017), 1–4.

[88] Watanabe, K., Oka, M. and Mori, H. 2019. Feedback Control to Target Joints Angle in Middle Finger PIP and MP Joint Using Functional Electrical Stimulation. *Human Interface and the Management of Information. Information in Intelligent Systems* (Cham, 2019), 440–454.

[89] Watanabe, T., Iibuchi, K., Kurosawa, K. and Hoshimiya, N. 2003. A method of multichannel PID control of two-degree-of-freedom wrist joint movements by functional electrical stimulation. *Systems and Computers in Japan*. 34, 5 (2003), 25–36. DOI:https://doi.org/10.1002/scj.10298.

[90] Widjaja, F., Shee, C.Y., Au, W.L., Poignet, P. and Ang, W.T. 2011. Using electromechanical delay for real-time anti-phase tremor attenuation system using Functional Electrical Stimulation. *2011 IEEE International Conference on Robotics and Automation* (May 2011), 3694–3699.

[91] Yu-Luen Chen, Weoi-Luen Chen, Chin-Chih Hsiao, Te-Son Kuo, and Jin-Shin Lai 2005. Development of the FES system with neural network + PID controller for the stroke. *2005 IEEE International Symposium on Circuits and Systems* (May 2005), 5119-5121 Vol. 5.

[92] Zubrycki, I. and Granosik, G. 2017. Novel Haptic Device Using Jamming Principle for Providing Kinaesthetic Feedback in Glove-Based Control Interface. *Journal of Intelligent & Robotic Systems*. 85, 3 (Mar. 2017), 413–429. DOI:https://doi.org/10.1007/s10846-016-0392-6.