Evaluation of a Body-Conforming Electrode for Functional Ultrasound Compatible Electrical Stimulation

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Abstract—Functional electrical stimulation (FES) is a neurorehabilitation modality that helps improve the size and strength of atrophied muscles after paralysis and reduce spasticity. However, FES causes muscle fatigue, and monitoring the patient's response to avert muscle fatigue caused by FES becomes imperative. Ultrasound (US) imaging can elucidate valuable information about muscle thickness, stiffness, force, and fatigue in the muscles. However, if the FES electrodes do not match the acoustic properties of the surrounding tissues, it could result in the interference with other electronic devices, potentially affecting both US imaging and the performance of the FES system. This study evaluates a specially designed body-conforming FES customized electrode that is intended to be compatible with US. The electrode is made with silver-nanowires/ Polydimethylsiloxane (AgNW/PDMS). The performance of the body-conforming customized FES electrode was demonstrated in parallel to that of the commercial hydrogel electrode. Moreover, compatibility with US was established through tests employing a 3.5 MHz single-element US transducer. The customized FES electrode exhibited a comparably minor variance (< 8%) relative to the commercial hydrogel electrode.

Keywords—Body-Conforming Ultrasound Compatible Electrode, Functional Electrical Stimulation (FES), Silver-Nanowires (AgNW) Electrode

I. Introduction

Spinal cord injury (SCI) and stroke are major neurological incidents leading to paralysis. Referring to the Traumatic Spinal Cord Injury facts and figures at a glance 2023 from National Spinal Cord Injury Statistical Center, the United States witnesses approximately 18,000 new patients diagnosed with SCI annually, and a total of 302,000 people with SCI are estimated to live in the United States. According to the American Stroke Association, stroke is the fifth fatal cause of death and disability in the United States. The physical impairment of one side of the body is a common consequence of stroke. The potential for recovering the mobility of the impaired body usually diminishes six months after having a stroke [1]. Substantially decreased quality of life motivates these individuals to seek rehabilitation interventions that ameliorate the symptoms of paralysis [1].

Functional electrical stimulation (FES) is a therapeutic application used in neurorehabilitation to facilitate the functional movement of a paralyzed limb. However, it is limited to use FES alone due to the rapid onset of muscle fatigue, resulting in swift loss of FES-elicited muscle force [2 – 5]. Sonomyography using ultrasound (US) imaging can evaluate muscle fatigue induced by the FES. Ultrasound imaging can penetrate deep into muscle to detect the activities, unlike other sensing modalities, e.g., surface electromyography (sEMG), which has a low signal-to-noise ratio (SNR) and an inability to monitor deeper muscles reliably [6, 7]. US offers relatively high resolution imaging and therefore can provide real time monitoring of muscles activities during FES stimulation [8]. In addition, since US imaging yields information about location rather than force, it cannot induce muscular fatigue, which bestows another benefit to its usage [9]. Given the merits of US imaging, it motivates us to develop an acoustically matched FES electrode, enabling precise muscle monitoring in the specific area of FES implementation.

Given these considerations, we designed and fabricated silver-nanowires/Polydimethylsiloxane (AgNW/PDMS) electrodes customized for ultrasound compatibility. The inclusion of PDMS serves to mitigate electromagnetic noise that arise from FES without the distortion of US imaging. A validation test was conducted to evaluate the effectiveness of the customized FES electrode by comparing it with commercial hydrogel FES electrodes. By measuring the wrist bend angle caused by wrist extensor stimulation, the performance of both FES electrodes was compared with different stimulation parameters: current amplitude, frequency, and pulse width. Furthermore, a 3.5 MHz single element ultrasound transducer was used for assessing the feasibility of the ultrasound compatible AgNW/PDMS customized electrode in this work. Since current commercially available FES electrodes can be hinderance for on-the-spot monitoring of muscle activities, this strategic integration was applied to validate the electrode's capacity to enable the transmission of ultrasound waves even when functional electrical stimulation (FES) is active, all the while circumventing significant insertion loss.

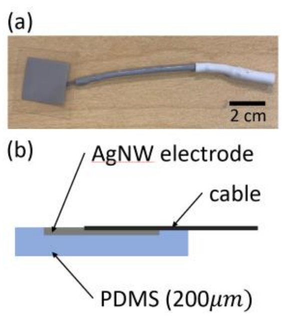

Fig. 1. AgNW/PDMS customized FES electrode (a) and the schematic of the structure of the electrode (b).

II. Methods

A. Ultrasound Compatible FES Electrode Design and Fabrication

Silver-nanowires (AgNWs) were synthesized by a modified polyol process [10, 11]. Next, AgNWs were drop-cast onto a precleaned glass slide with Kapton tape as a sacrificial mask. AgNWs were thermal annealed to fuse the AgNW junctions and to uniform and conductive network. The thickness of AgNW film ranged from one to several microns. The resulting AgNW film was patterned using a laser cutter (Universal Laser System VLS 6.60 Laser) to dimensions of 2 cm by 2 cm. Degassed liquid PDMS (sylgard 184, Dow Corning) was spin-coated on the AgNW film at 400 rpm for 30 seconds and cured at 50 °C for 12 h. Once cured, AgNWs were embedded just below the PDMS surface to form AgNW/PDMS electrode. A commercial cable (PALS® Electrodes, Axelgaard) was attached to the AgNW/PDMS customized electrode for connecting it to the current-controlled stimulator (Rehastim1, HASOMED GmbH). Fig. 1. illustrates the AgNW/PDMS customized FES electrode (a) and the schematic representation of the electrode's structure (b).

B. Evaluation of the Effectiveness of Ultrasound Compatible FES Electrode

A commercial hydrogel electrode (PALS® Electrodes, Axelgaard) was chosen for comparison study. To mitigate the potential impact of muscle fatigue induced by FES stimulation, a time interval was introduced between testing sessions involving the commercial electrode and AgNW/PDMS customized electrodes. This interval is a crucial consideration, as it allows for the dissipation of any residual muscle fatigue effects from the previous stimulation. By introducing this time gap, we aim to ensure that each testing session commences with a relatively consistent baseline, thereby minimizing the potential effects of cumulative fatigue on the experimental outcomes. The experimental setup for the validation test is depicted schematically in Fig. 2. (a). The validation tests were conducted on the right forearm. The FES electrodes were placed atop the forearm, specifically at 5 cm from the lateral condyle, as well as the tendinous region, 5 cm from the carpals, as illustrated in Fig. 2. (b – c). Stimulation was applied to the forearm extensor to induce the wrist extension.

To quantify the wrist bend angle resulting from FES stimulation, an inertial measurement unit (IMU) was used. Relaxed status corresponded to a wrist bend angle of 0 °. The angular change of the wrist, prompted by the FES stimulation delivered through the current-controlled stimulator (Rehastim1, HASOMED GmbH) was measured. This study aimed to assess the effects of distinct stimulation parameters, including current amplitude (5,8, and 11 mA), frequency (20,30, and 40 Hz), and pulse duty cycle (20%, 40%, and 60%). Each FES stimulation lasted for 3 seconds and was repeated four times.

C. Experimental Setup

The schematic representation of the experimental setup for feasibility test is illustrated in Fig. 2. (d). A 3.5 MHz single-element ultrasound transducer was employed in the feasibility test to estimate the insertion loss of both the customized and commercial electrodes. The single-element ultrasound transducer was placed directly on top of the electrodes in a vertical orientation to determine the acoustic compatibility of customized electrode, as depicted in Fig. 2. (e – f). A pulser/receiver (5077 PR, Olympus, WA, USA) was connected to the transducer, which operated at a pulse repetition frequency (PRF) of 200 Hz and input voltage of 200 V. A bandpass filter was set within the range of 1 to 10 MHz. An oscilloscope (DSO7104B, Agilent Technologies, Santa Clara, CA, USA) was utilized to display the Radiofrequency (RF) signal.

For stimulation of the wrist extensor, the FES electrodes were placed on both the top of the forearm and the tendinous region. These electrodes were connected to a current-controlled stimulator. We selected parameters of 11 mA for current amplitude, 30 Hz for frequency, and 40 % for the pulse duty cycle.

Initially, B-mode imaging of the target muscle was collected using a Clarius ultrasound probe (Vancouver, Canada) to establish a reference for the A-mode echo signal of the single element ultrasound transducer. Time of flight (TOF) measurements were calculated and compared with B-mode imaging data. Subsequently, echo signals were obtained with and without the customized FES electrode to ascertain insertion loss. Lastly, the single element transducer was positioned above both the customized and commercial electrodes to evaluate the feasibility of collecting echo signals.

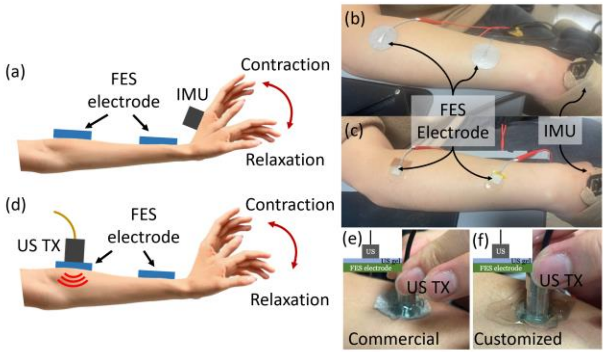

Fig. 2. experimental setup for evaluation of the effectiveness of the customized FES electrode (a – c). While wrist extensor is contracted by FES stimulation, IMU tracks the bend angle of wrist (a). experimental setup with commercial FES electrode (b), and the customized FES electrode (c). The experimental setup for feasibility test of the customized FES electrode (d – f). 3.5 MHz single element ultrasound (US) transducer collects A-mode US signal of muscle to see if the electrode is ultrasound compatible (d). feasibility test of commercial FES electrode for US transducer placed on top (e), and customized FES electrode(f)

III. Results

A. Effectiveness of Ultrasound Compatible FES Electrode

To evaluate the effectiveness of the customized FES electrode, a comparative test was conducted against the commercial FES electrode. Fig. 3. shows the electrical stimulation effect of the customized FES electrode on the wrist extensor, juxtaposed with the performance of commercial FES electrode. Various stimulation parameters: current amplitude, frequency, and pulse duty cycle of electrical stimulation were adjusted for the electrical stimulation. Notably, 5 mA of current amplitude was found to be the sub-threshold for both electrodes. In terms of performance, the customized FES electrode exhibited a comparably minor variance (< 8%) relative to the commercial hydrogel electrode.

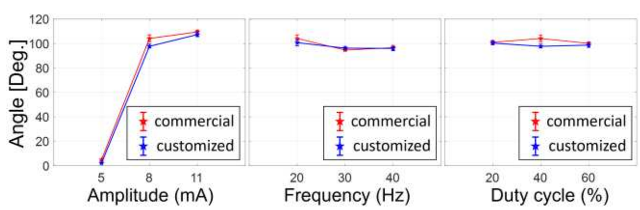

Fig. 3. Evaluation of the electrical stimulation effect of the customized FES electrode on the wrist extensor response in comparison to the performance of the commercial FES electrode with different stimulation parameters: current amplitude (5–11 mA), frequency (20–40 Hz), and pulse duty cycle (20–40 %) of electrical stimulation

B. Feasibility of Ultrasound Compatible FES Electrode

Initially, B-mode imaing was captured as reference for the A-mode ultrasound signals acquired using a single element ultrasound transdcuer. Fig. 4. (a) illustrates ultrasound imaging of the target muscle during states of relaxation and contraction of the wrist extensor. The muscle layers (labeled i and ii) constitute the regions of interest. In the relaxed state, these layers of muscle were situated approximately 1.5 cm and 2.1 cm beneath the skin surface, respectively. Following FES stimulation, the layers moved around 1.8 cm and 2.3 cm deeper from the skin, respectively. In fig. 4. (b), the first echo occured at 20.6 μs, corresponding to a distance of 1.58 cm, as calculated using time of flight. During muscle contraction, the subsequent echo at 27.25 μs corresponds to 2.1 cm in the relaxed state. During muscle contraction, the echo (labeled i) appeared at 23.75 μs, signifying a depth of 1.83 cm. The subsequent echo (labeled ii) at 30.3 μs indicated a distance of 2.3 cm, closely aligned with the location of muscle layers observed in B-mode imaging.

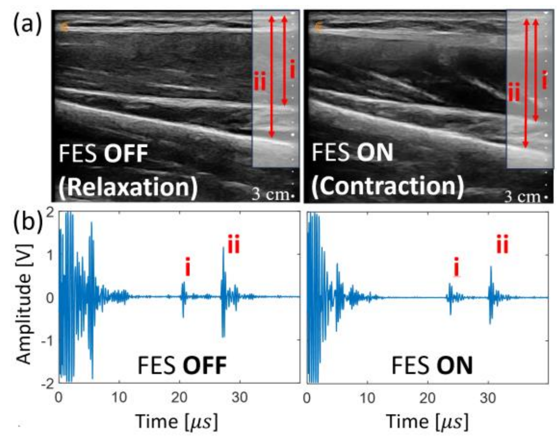

Fig. 4. B-mode ultrasound imaging of target layers of the muscle (i & ii) as reference at the status of relaxation and contraction of the wrist extensor (a). A-mode echo signal of the single element ultrasound transducer from muscle layers i & ii to see if time of flight (TOF) was matched with layers of muscle presented in B-mode ultrasound imgaing (b).

Next, to see the insertion loss of customized FES electrode (depicted in Fig. 5. (a)), changes in peak-to-peak voltage (V_{pp}) were measured based on the presense or absence of the customized FES electrode. Without the customized FES electrode, the V_{pp} value of the first and second echo signal (labeled i and ii) were measured to be 1.48 V and 2.23 V, respectively. When the customized FES electrode was positioned vertically on the front of the transducer, the V_{pp} value of the the first and second echoes (labeled i and ii) from the muscle layers were recorded as 0.83 V and 2.1 V. Although the first echo experienced a reduction of 44 %, the following echo decreased by 6 % in comparison to the V_{pp} measurement taken without the customized FES electrode.

Finally, the customized electrode was compared with the commercial FES electrode. As shown in Fig. 5. (b), the echo signal from the muscle layers was successfully captured in the presense of the customized electrode. However, the commercial electrode cannot yield ultrasound signal from the muscle layers. The findings demonstrated the customized electrode was compatible with ultrasound, allowing the transducer to effectively capture signals on the area of FES stimulation.

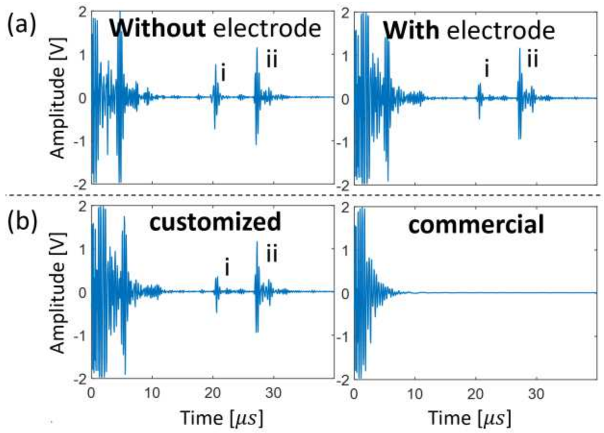

Fig. 5. Peak-to-peak voltage (V_{pp}) changes were measured to see the insertion loss depending on the presense of the customized FES electrode (a). the echo signals from the layers of the muscle (i & ii) were collected to compare the effects of the presense of the customized and commercial FES electrode on the front of ultrasound transducer (b).

IV. Discussion

T The tests unveiled that the AgNW/PDMS customized electrode can be ultrasound compatible. Unlike commercial FES electrode, PDMS on the customized FES electrode effectively shielded the EMI when the ultrasound transducer was vertically positioned on the electrode. We applied Tensive® conductive adhesive gel (Parker Laboratories, Inc., New Jersey) on the AgNW side of the electrode. Although the gel helped to secure the electrode to the target area, its application was messy and was difficult to clean post experiment. It also irritated the patients' skin upon removal.

The rigid single element ultrasound transducer should require exertion of pressure to elicit echo signals from the target muscle even amidst FES stimulation. The pressure onto the skin could potentially affect the alignment of the FES electrodes with the intended stimulated area, despite the usage of the adhesive gel. Referring to Fig. 4. (b), the reduction in peak-to-peak voltage by 44% for the first echo and 6 % for the second echo could be attributed to the alignment of the transducer with each muscle layer. Thus, this highlights the potential necessity for a wearable single element ultrasound transducer and arrays that not only eliminate the need for pressure on the transducer to collect data, but also mitigate concerns regarding misplacement of both the FES electrode and the ultrasound transducer.

V. Conclusion

This study introduced an AgNW/PDMS customized FES electrode designed to be ultrasound compatible, and the effectiveness of the electrical stimulation electrode was evaluated across various stimulation parameters. The assessment revealed a performance akin to that of the commercial hydrogel electrodes. Importantly, the customized FES electrode, positioned vertically beneath the transducer, was acoustically matched so that the ultrasound transducer can capture the echo signal from the forearm muscle layers with minimal insertion loss. The findings strongly suggest the promising prospects of an ultrasound compatible FES electrode, enabling precise monitoring of muscle activities in the intended region targeted by FES stimulation.

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References

[1] C. Marquez-Chin and M. R. Popovic, "Functional electrical stimulation therapy for restoration of motor function after spinal cord injury and stroke: a review," BioMedical Engineering OnLine, vol. 19, no. 1, May 2020

[2] C. Marquez-Chin and M. R. Popovic, "Functional electrical stimulation therapy for restoration of motor function after spinal cord injury and stroke: a review," BioMedical Engineering OnLine, vol. 19, no. 1, May 2020

[3] R. Martin, C. Sadowsky, K. Obst, B. Meyer, and J. McDonald, "Functional Electrical Stimulation in Spinal Cord Injury: From Theory to Practice," Topics in Spinal Cord Injury Rehabilitation, vol. 18, no. 1, pp. 28–33, Jan. 2012

[4] Z. Sheng, A. Iyer, Z. Sun, and N. Sharma, "A Hybrid Knee Exoskeleton Using Real-Time Ultrasound-Based Muscle Fatigue Assessment," vol. 27, no. 4, pp. 1854–1862, Aug. 2022

[5] Q. Zhang, A. Iyer, K. Lambeth, K. Kim, and N. Sharma, "Ultrasound Echogenicity as an Indicator of Muscle Fatigue during Functional Electrical Stimulation," Sensors, vol. 22, no. 1, p. 335, Jan. 2022

[6] X. Sun, Y. Li, and H. Liu, "Muscle fatigue assessment using one-channel single-element ultrasound transducer," May 2017

[7] P. Li, X. Yang, G. Yin, and J. Guo, "Skeletal Muscle Fatigue State Evaluation with Ultrasound Image Entropy," Ultrasonic Imaging, vol. 42, no. 6, pp. 235–244, Aug. 2020

[8] Q. Zhang, A. Iyer, K. Lambeth, K. Kim, and N. Sharma, "Ultrasound Echogenicity as an Indicator of Muscle Fatigue during Functional Electrical Stimulation," Sensors, vol. 22, no. 1, p. 335, Jan. 2022

[9] X. Xue et al., "Development of a Wearable Ultrasound Transducer for Sensing Muscle Activities in Assistive Robotics Applications," Biosensors, vol. 13, no. 1, p. 134, Jan. 2023

[10] F. Xu and Y. Zhu, "Highly Conductive and Stretchable Silver Nanowire Conductors," Advanced Materials, vol. 24, no. 37, pp. 5117–5122, Jul. 2012

[11] D. Shukla, Y. Liu, and Y. Zhu, "Eco-friendly Screen Printing of Silver Nanowires for Flexible and Stretchable Electronics," Nanoscale, 2023