Body-conforming Multi-modal Sensing with Ultrasound-compatible Surface Electromyography Sensors and a Wearable Ultrasound Array

Sunho Moon
The Department of Mechanical and Aerospace Engineering
North Carolina State University
Raleigh, NC, USA
smoon4@ncsu.edu

Xiangming Xue
The joint Department of Biomedical Engineering
North Carolina State University
Raleigh, NC, USA
xxue5@ncsu.edu

Vidisha Ganesh
The joint Department of Biomedical Engineering
North Carolina State University
Raleigh, NC, USA
vganesh3@ncsu.edu

Darpan Shukla
The Department of Mechanical and Aerospace Engineering
North Carolina State University
Raleigh, NC, USA
dshukla2@ncsu.edu

Yong Zhu
The Department of Mechanical and Aerospace Engineering
North Carolina State University
Raleigh, NC, USA
yzhu7@ncsu.edu

Nitin Sharma
The joint Department of Biomedical Engineering
North Carolina State University
Raleigh, NC, USA
nsharm23@ncsu.edu

Xiaoning Jiang
The Department of Mechanical and Aerospace Engineering
North Carolina State University
Raleigh, North Carolina, USA
xjiang5@ncsu.edu

Abstract—Surface electromyography (sEMG) is a widely used non-invasive diagnostic modality in rehabilitation, designed to detect muscle activation in response to neural signals at the skin's surface. However, sEMG has limitations, including its inability to accurately measure deeper muscle activation, low spatial resolution, susceptibility to noise interference, and crosstalk from neighboring muscles. In contrast, ultrasound (US) imaging provides valuable insights into muscle thickness, stiffness, force, and fatigue with high spatial resolution and detection accuracy. Furthermore, the US can assess both superficial and deep muscle layers, addressing the shortcomings of sEMG. The integration of sEMG with US can provide comprehensive insights into both electrophysiological and morphological activities of muscle simultaneously. To facilitate this, we developed ultrasound-compatible EMG (US-EMG) sensors and a 64-element wearable ultrasound (WUS) array. The US-EMG sensor was fabricated using silver-nanowires and polydimethylsiloxane (AgNWs/PDMS) composite. The fabricated US-EMG sensor performed comparably to the commercial EMG electrodes. Next, The WUS was electrically and acoustically characterized, demonstrating uniformity across the array. Successful acquisition of EMG signals and B-mode US imaging without significant signal loss was achieved during the vertical integration of the WUS array with the US-EMG sensors. On the contrary, the commercial EMG sensors did not allow for clear US imaging. This multi-modal sensing device shows great potential for enhancing motion prediction and muscle fatigue detection.

Keywords—Body-Conforming Ultrasound-Compatible Electrode, surface Electromyography (sEMG), Silver-Nanowires (AgNW) electrode, Wearable Ultrasound Array Transducer

I. Introduction

According to the World Health Organization (WHO), 15 million people suffer strokes worldwide annually. Of these, 5 million are left permanently disabled. The most common deficit after stroke hemiparesis of the contralateral upper limb, with more than 80% of stroke patients experiencing this condition acutely [1]. About half of all stroke survivors report challenges in upper extremity motor function six months post-stroke, which can include hemiparesis, spasticity, co-contraction, pain, or other limitations which impact quality of life [2].

Several wearable sensor options exist for monitoring recovery in stroke survivors. Surface electromyography (sEMG) serves as a non-invasive diagnostic method in neuromuscular interface to measure muscle response to neural electrical signals [3]. EMG translates electrical signals during muscle contractions via small electrodes attached to the skin's surface. Despite its widespread use, sEMG faces challenges such as limited depth of sensing, and interference from adjacent muscles. Conversely, B-mode Ultrasound (US) imaging presents an alternative non-invasive approach to monitor muscle activity, particularly in deep muscles, while minimizing interference from adjacent muscle activity. Nevertheless, using US imaging solely poses challenges in accurately interpreting motion intention [4, 5]. By harnessing the complementary nature of sEMG, which provides information about electrical muscle activity, and US imaging which offers insight into muscle contractility, a dual-modal sensing device has great potential for motion prediction and muscle fatigue detection in applications of post-stroke monitoring and recovery [6].

There is a notable gap in studies exploring the effectiveness of WUS combined with US-EMG sensors. Integrating these technologies could enhance the simultaneous monitoring of electrical muscle activity and muscle morphology, providing a more comprehensive approach to assessing motor function recovery in stroke patients. In this study, we present multi-modal sensing device fusing ultrasound-compatible EMG (US-EMG) sensors and wearable ultrasound (WUS) array to monitor the muscle activity, enabling to provide electrophysiological information from EMG and morphological information of muscle simultaneously. We demonstrate the effectiveness of US-EMG sensors against the commercial EMG sensors and the

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feasibility of integrating WUS array and US-EMG sensors in this work.

II. Methods

A. Design and Fabrication of Ultrasound-compatible Electromyography Sensor

To fabricate body-conforming electrodes [7, 8], silver-nanowires (AgNWs) were synthesized by a modified polyol process. The AgNWs were drop-cast onto a pre-cleaned glass slide with CO_2 laser patterned (Universal Laser System VLS 6.60 Laser) Kapton tape as a sacrificial mask. The AgNWs were thermally annealed to fuse the AgNW junctions, forming a uniform and conductive network. The thickness and width of AgNW film ranged from a few microns to several microns and 20 mm diameter, respectively. Degassed liquid polydimethylsiloxane (PDMS, sylgard 184, Dow Corning) was spin-coated onto the AgNW film at 400 rpm for 30 seconds and cured at 50 °C for 12 hours. Upon curing, the AgNWs were embedded just beneath the surface of PDMS, forming a surface-embedded AgNW/PDMS electrode. A commercial cable with 1.5 mm safety connectors (Sticky Pad™ EMG electrode, Rhythmlink®) was then attached to the AgNW/PDMS electrode to enable connection to a neural recording system (Tucker-Davis Technologies) for EMG signal acquirement. Figure 1. illustrates the AgNW/PDMS customized EMG sensors (a) and the side view of schematic representation of the sensor's structure (b).

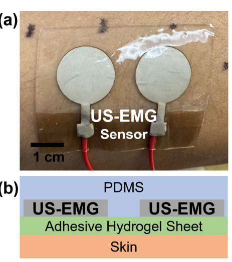

Fig. 1. (a) the US-EMG sensors fabricated with AgNW/PDMS, and (b) the side view of schematic representation of the sensor's structure

B. Design and Fabrication of Wearable Ultrasound Array

The element configuration of wearable ultrasound (WUS) array was designed with Krimboltz, Leedom, and Mattaei (KLM) equivalent circuit model to simulate its electrical and acoustical properties. A center frequency of 6 MHz was selected to achieve the sufficient penetration depth of US wave for imaging the muscle activity. The simulation of the beam focusing and beam steering profiles of WUS, consisting of 64 elements, was also conducted to explore the capabilities of muscle imaging with the appropriate resolution and field of view. To achieve the optimal properties of muscle imaging, a piezoelectric composite material was selected to have high electromechanical coupling factor and acoustic impedance below 20 Mrayls. Additionally, the pitch size of 160 µm was chosen as less than one wavelength to ensure high-resolution ultrasound imaging performance by minimizing the effects of grating lobes. Table 1 summarizes the design parameters.

PZT-5H/Epoxy piezo-composite elements were fabricated using a dicing machine (DAD322, DISCO) with the blade thickness of 30 µm, which was kerf width. Kerf was filled with epoxy resin (Epo-Tek 301, Epoxy Technologies Inc.). It was lapped down to reach a center frequency of 6 MHz. A backing layer composed of E-solder 3022 (centrifuged) was mounted to the elements of WUS. A customized flex printed circuit (FPC) was bonded along the elements, followed by the connection to a printed circuit board (PCB) for the incorporation of the Verasonics ultrasound research system. A matching layer of Al_2O_2/Epoxy was affixed to the surface for enhanced wave transfer from the transducer to the tissue. 3D printed housing and wearable band were installed.

TABLE I. Designed properties of wearble ultrasound array

Frequency	6 MHz
Aperture size (mm)	10.2 × 9
Element Number	64
Pitch	160 µm
Kerf	30 µm

C. Characterization of Wearable Ultrasound Array

Comprehensive electrical and acoustical characterization was performed on the WUS array before its deployment. Electrical impedance, phase angle spectrum, capacitance, and dielectric loss was measured using an impedance analyzer (Agilent 4294A, Agilent Technologies). A pulse-echo testing was performed to figure acoustic properties including sensitivity, center frequency, and bandwidth. Each element was activated using a pulser/receiver (Olympus 5077 PR, Olympus Corp.) with a pulse repetition frequency (PRF) of 200 Hz and a pulse energy of 1 µJ. It is set through a bandpass filter ranging from 3 to 20 MHz for capturing echo signals. The steel bar was used as reflector to collect the reflected echo signals. An oscilloscope (DSO7014B, Agilent Technologies) was used to display and save the resulting raw radiofrequency (RF) data. Given the saved RF data, acoustic properties of each element of WUS were analyzed using MATLAB software.

D. Experimental setup for a Comparative Study of EMG signals between US-EMG sensors and commercial EMG sensors

A commercial EMG sensor (Sticky Pad™ EMG electrode, Rhythmlink®) was chosen for comparison study. The experimental setup for the comparison test is depicted schematically in figure 2. For the validation test, a commercial EMG electrode (Sticky Pad™ EMG electrode, Rhythemlink®) with a 20 mm diameter was selected. Two electrodes were closely positioned on forearm muscle (5 cm from the lateral condyle) with interelectrode distance (IED) of 30 mm. Other electrodes on the abdomen for reference and ground connection, respectively. This setup facilitated the isolated depiction of wrist extension. Volitional wrist movements ranging from 0 to 60 degrees were performed, with three trials conducted for each test. EMG signals generated by wrist movement were collected using an EMG system (Tucker-Davis

Technologies) for both the commercial EMG and the US-EMG sensors. The collected raw EMG data was processed using a 10 – 50 Hz bandpass filter, rectification, and smoothing.

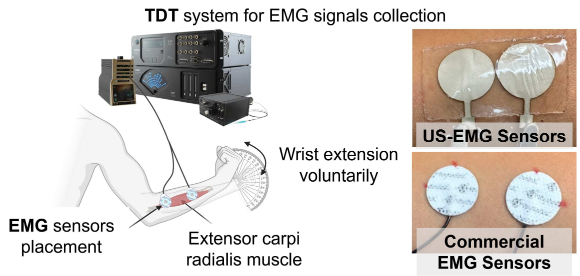

Fig. 2. Comparison Study of EMG Signal between US-EMG sensors and Commercial EMG Sensors

E. *Experimental setup for Integration Feasibilty Tests of Wearable Ultrasound Array with US-EMG sensors and Commercial EMG Sensors*

A programmable US research system (Vantage 256, Verasonics Inc.) was used to see the integration feasibility of WUS with both US-EMG sensors and commercial EMG sensors. WUS was controlled by the verasoncis system for signal transmission and reception to reconstruct the real time B-mode images of the targeted muscle using MATLAB software (R2023a, MathWorks). WUS was connected to the verasonics system and placed on top of the EMG sensors. Wearable band allowed the WUS to hold tight on the EMG sensors to monitor the targeted muscle where EMG signals are being recorded. While EMG signals recorded, the real time B-mode US imaging of targeted muscle contraction can be obtained through the WUS array. The corresponding experimental setup is illustrated in figure 3.

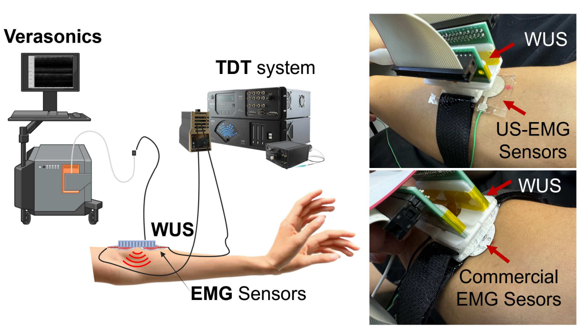

Fig. 3. WUS Integration Tests with US-EMG Sensors and Commercial EMG Sensors

III. Results

A. *Evaluation of effectiveness of US-EMG sensors*

To evaluate the effectiveness of the custom-made US-EMG electrode, a comparative test was conducted against the commercial EMG electrode. Figure 4 shows the EMG signals created by the wrist extension, juxtaposed with the performance of commercial EMG electrode. In terms of performance, the mean, peak, and RMS value of EMG signals collected from US-EMG electrode were 75.53 µV, 338.26 µV, and 116.42 µV, respectively. Corresponding values from commercial EMG electrode were 54.74 µV, 282.78 µV, and 86.55 µV.

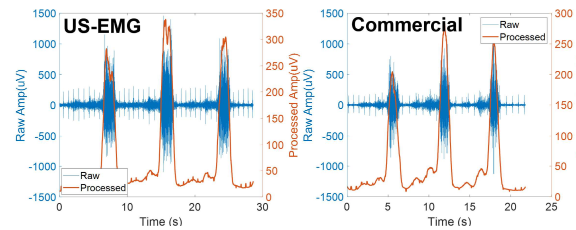

Fig. 4. Evaluation of effectiveness of US-EMG electrode

B. *Characterization of Wearable Ultrasound Array*

The electrical and acoustic performance was evaluated, and the representative electrical impedance magnitude and its corresponding phase angle spectrum as well as impulse response and its frequency spectrum are shown in figure 5. Electrical performance of each element was analyzed for characterizing WUS array. Figure 6 (a–e) shows the measured electrical characteristics of each element of WUS including electrical impedance magnitude, phase angle spectrum, resonant frequency, capacitance, and dielectric loss. The average and deviation values of 209.30 ± 0.023 *Ω*, -42.20 ± 2.59 °, 6.09 ± 0.25 MHz, 91.87 ± 5.47 pF, and *21.56 ± 1.60 mU* respectively, showing high performance consistency across the elements of WUS. The acoustic characteristics of elements were also obtained by pulse-echo response tests and organized, including measured peak-to-peak voltage, -6 dB bandwidth, and center frequency of elements. The average and deviation values of 20.29 ± 0.26 mV, 46.23 ± 9.42 %, and 5.88 ± 0.55 MHz, respectively, visualizing the uniformity for all array elements as shown in figure 6 (f–h).

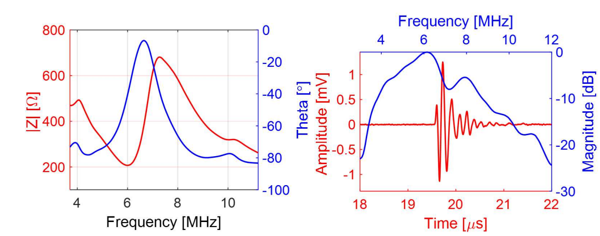

Fig. 5. A representative electrical impedance magnitude, its corresponding phase angle spectrum as well as impulse response and its frequency spectrum

C. *Feasibiilty of integrating a Wearable Ultrasound Array with US-EMG sensors*

The real time B-mode imaging of targeted muscle contraction using WUS and Verasonics US array was shown in figure 7. WUS was used to collect the US images by being placed on the skin (figure 7a), top of US-EMG sensors (figure 7b) and the commercial EMG sensor (figure 7c). The real time B-mode US imaging was performed to monitor clearly the targeted muscle activity even though US-EMG electrode was placed in the path of US wave penetration. There was no significant signal loss in comparison to the B-mode imaging from the experimental condition of no electrode. However, half of the US imaging was obscured by the commercial EMG sensor as the material did not allow acoustic wave penetration.

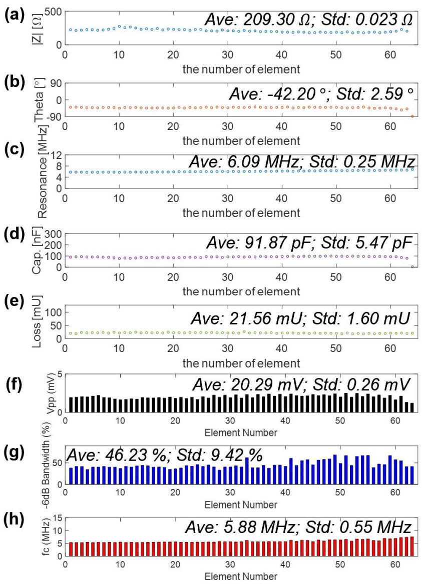

Fig. 6. Chracterization of WUS, showing consistency and uniformity of elements of WUS

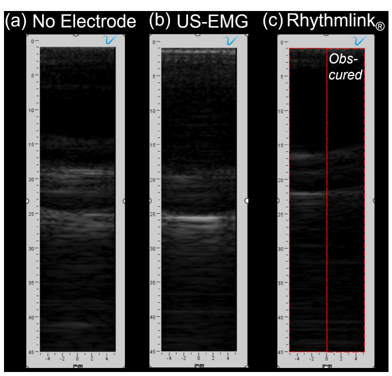

Fig. 7. Integration of WUS with EMG sensors (B-mode acquirement)

IV. Futer works & Discussion

Future work will focus on a flexible and wearable ultrasound (fWUS) array transducer, consisting of 128 elements. Fabrication processes will be modified by replacing epoxy resin with PDMS as the filler material. To make it fully body-conforming, the fWUS array may need a dry or semi-dry coupling medium instead of using ultrasound conductive gel. Once the gel connects skin to the US array, even a subtle amount of electrical noise may hinder the EMG signals due to the susceptibility of electrical noise of EMG sensors. Thus, this highlights the potential necessity for the multi-modal sensing device that mitigate concerns regarding the interference.

V. Conclusion

This study introduced a multi-modal sensing device combining a US-compatible EMG sensor and WUS array. The effectiveness of the proposed US-EMG sensor was demonstrated by comparison against a commercial EMG sensor. Moreover, the feasibility of integrating US-EMG sensors with WUS was shown by constructing the real time B-mode images of muscle activity during which EMG signals were recorded. However, the commercial EMG sensors did not allow for US imaging due to the limitations of US wave penetration through the sensors. In conclusion, our dual-modal approach combining US-EMG sensors and WUS array provides valuable insights into electrical and morphological activities of muscle simultaneously, resulting in enhanced motion prediction and muscle fatigue detection.

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