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RESEARCH ARTICLE

Development of Consumer-Friendly Surface Electromyography System for Muscle Fatigue Detection

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This work involved human subjects or animals in its research. Approval of all ethical and experimental procedures and protocols was granted by the Human Research Ethics Committee at Kanagawa University in 2022, and performed in line with the Declaration of Helsinki.

ABSTRACT In this study, a low-cost, wireless, and smartphone-controlled surface electromyography (EMG) system was designed and developed for consumers, and the recorded EMG signals were evaluated against a reference laboratory EMG system during fatiguing contraction. Using commercially available inexpensive components, the components of the EMG signal-acquisition circuit were optimized, and a microcontroller was combined with a Bluetooth module. The EMG signals were then converted from analog to digital signals and transmitted to a smartphone via Bluetooth serial communication. EMG signals from the biceps brachii of six healthy subjects were recorded separately using two EMG systems during sustained submaximal isometric contraction until the endurance limit was reached. The root mean square (RMS) and mean power frequency (MPF) of the EMG signals were calculated. The results indicated that both the EMG systems exhibited a characteristic progressive increase in EMG_{RMS} and decrease in EMG_{MPF} during sustained isometric contraction. The relative agreement between the two EMG systems, assessed by intraclass correlation coefficient (ICC), was excellent for EMG_{RMS} (ICC 0.933, $P < 0.001$) and moderate for EMG_{MPF} (ICC 0.662, $P = 0.049$). The cost of the sensor components in the hardware was ¥8,486 per unit. The proposed consumer-friendly EMG system, which is inexpensive and highly versatile in terms of wireless and smartphone accessibility, can detect the phenomenon associated with increased amplitude and low-frequency components during muscle fatigue contraction with a magnitude similar to that of the commercially available laboratory EMG systems.

INDEX TERMS Amplitude, EMG, low-cost, smartphone, spectral analysis, sustained isometric contraction.

I. INTRODUCTION

Accurate muscle fatigue estimation is essential for various sports and rehabilitation. It also aids significantly in achieving efficient training and rehabilitation outcomes. Surface electromyography (EMG) is a technique used to estimate muscle fatigue. Using sustained submaximal isometric contractions that induce muscle fatigue, defined as a progressive exercise-induced decline in the force-generation capacity [1], EMG signals are expected to exhibit increased amplitude and decreased frequency [2]. For EMG systems to be useful in frequently detecting muscle fatigue, they must be highly versatile, cost-effective, non-invasive, and wireless for portability and ease of use.

Surface EMG systems have traditionally been used in laboratory environments. However, they are typically expensive and not portable; moreover, the electrode and amplifier module are often wired. Some commercial EMG devices are reportedly expensive, with costs ranging from approximately € (EUR) 15,000 to €20,000 (approximately ¥(JPY) 2,200,000 to ¥2,942,000) [3] or up to $ (USD) 20,000 (approximately ¥2,750,000) [4]. For this reason, in several

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studies, a low-cost EMG system has been developed. For example, Bawa [5] developed an EMG system using commercially available specialized modules, MyoWare and Arduino, for EMG measurement at a total cost of $150 (approximately ¥20,500). Yassin [6] developed EMG system using commercially available microcontrollers (MCUs) and Bluetooth chips without using specialized modules for EMG measurement, at a total cost of $69.99 (approximately ¥9,500). Although these two studies do not mention all the components used, the total costs are roughly equivalent to that in other studies listing all components: Fortune et al. [7] at $112 (approximately ¥15,000 and McKenzie et al. [8] at $145 (approximately ¥19,500). To the best of our knowledge, only two studies has attempted to evaluate the variations in the EMG signals associated with muscle fatigue using low-cost EMG system [3], [5]. To analyze the amplitude and frequency of EMG signals, del Toro et al. [3] used low-cost modules combinations of MyoWare sensor chip and Arduino board. However, these two studies using developed low-cost EMG system have qualitatively compared the amplitude and frequency of EMG signals during fatiguing contraction against a laboratory-standard EMG system, which is yet to be comprehensively validated using the degree of agreement between two EMG systems. In addition, the EMG system proposed in these studies were not versatile as the connection between electrode and board were wired, and they required postprocessing using MATLAB (a highly specialized analysis software) on a personal computer (PC) having low portability. Therefore, unless the drawbacks of not only the high cost but also the completely wired connection, postprocessing, and portability are resolved, the general population is unlikely to use them on a daily basis.

Therefore, a consumer-friendly surface EMG system that consumers can use on a daily basis was designed and developed in this study to assess muscle fatigue. This study is the first attempt in developing a highly versatile low-cost, fully wireless, and smartphone-accessible EMG system. Low cost was achieved by using only commercially available inexpensive components. Furthermore, the EMG signals recorded during fatiguing contractions are compared and validated against a reference laboratory EMG system.

II. LITERATURE REVIEW

The development of low-cost EMG systems can be divided into two categories: custom-made and commercial products. In this review, commercial products defined as those that the plug-in-play EMG device such as Trigno as well as dedicated components specialized for the EMG measurement such as the MyoWare and BITalino while custom-made products as those manufactured by combining components that are not modules specialized for the EMG measurement. In this section, we have reviewed recent studies that developed devices aimed at measuring EMG signals noninvasively, focusing on aspects related to versatility, such as wireless connection, EMG signal post-processing, and cost.

A. CUSTOM-MADE PRODUCTS

The biggest advantage of custom-made products is cost effectiveness. In previous studies, clarity regarding the cost of all components used to fabricate the EMG systems is very limited. Fortune et al. [7] developed an open-source active EMG circuit using thirteen components, and the total cost of hardware was $112 (approximately ¥15,000) for one and $750 (approximately ¥100,300) for ten. This device allowed a large full scale input range, low baseline noise, and good interference suppression, thus bridging the gap between high quality EMG and affordability. McKenzie et al. [8] designed, developed, and tested unique EMG, able to record EMG signal concurrently to electrical stimulation. This design used total 59 components for fabrication, and the total cost was $145 (approximately ¥19,500 for one and $548 (approximately ¥75,000) for five. Muraoka et al. [9] developed a low-cost EMG device comprising an EMG amplifier, a PC with a microphone port, an electrode, and their cables and reported the cost of the 11 components used to develop their system at ¥5,200. The cost of these custom-made products is clearly lower than that of the commercial products described below.

In contrast, most custom-made products utilize some type of lead wire (i.e., wired connection) from the electrode to the post-processing apparatus [7], [8], [9], [10], [11], [12], [13], [14], [15], [16], [17]. To the best of the authors' knowledge, none of these systems are completely wireless. Zhao et al. [17] developed a readout circuit for acquiring EMG signals with six surface electrodes, comprising of wired connections between the EMG acquisition circuit and electrodes as well as the EMG acquisition circuit and battery. Although Yang et al. [16] eliminated the wires by placing the substrate and electrodes in the same case, they used lead wires to connect the substrate to the reference electrode. This is a concern owing to its potential artifacts owing to cable movement as well as restriction of body movement and clothing during EMG measurement.

For EMG signal post-processing, several studies have utilized MATLAB or LabVIEW on a PC [7], [9], [10], [11], [12], [13], [14], [16], [17]. For example, Geethanjali and Ray [12] used a PC for processing the stored EMG data in MATLAB. One of the issues was that a power adapter was required to use the desktop PC, and specialized knowledge was needed to work with MATLAB and LabVIEW. Conversely, Yassin et al. [6] developed a handheld EMG biofeedback device comprising a physiological signal-acquisition circuit and a smartphone with a cloud server. The smartphone collected the EMG signals from the acquisition circuit and uploaded them to the cloud. Suzuki et al. [18] visualized the EMG wave form on a smartphone. Wu et al. [15] transmitted data using a Bluetooth low-energy module to a smartphone, which then used an infinite impulse response filter to remove noise and display the EMG wave form on the screen. However, to the best of the authors' knowledge, none of the aforementioned studies analyzed the EMG signals on the smartphone itself.

Finally, comparing the EMG signals measured by the custom-made products with those of a commercially available EMG system is essential. Although few studies have developed such systems, only three studies have quantitatively verified their validity with correlation coefficients of over 0.8 [6], [15], [16] representing moderate-to-high relative agreement between the custom-made and commercial EMG systems. For the former, the EMG signals must also be verified to be consistent with those of commercial products.

B. COMMERCIAL PRODUCTS

Two types of commercial products have been observed: systems with plug-in-play device and specialized board infrastructure for EMG measurement. BiostampRC [2], mDurance [19], and Duelite [20] have been primarily used in studies as systems with inexpensive plug-in-play device. Doheny et al. [2] used the BiostampRC, a low-power MCU that acquires signals from the EMG sensors as well as a 3-axis accelerometer and gyroscope. It then processed, sampled, and transmitted data to flash memory or broadcasted wirelessly via Bluetooth. The data collected during fatiguing contraction were then post-processed in MATLAB on a PC. While the BiostampRC was completely wireless, the mDurance and Duelite utilized a wired connection between the electrodes and bridge amplifier or probe for sampling. For post-processing, the mDurance Android mobile application receives the EMG data from the acquisition unit and transmits it to a cloud service. The other plug-in-play devices analyze the EMG signals via commercial programming software, such as MATLAB, necessitating programming skills and specific knowledge. The cost of these plug-in-play devices range from €538 (approximately ¥78,600) to ¥680,000.

For specialized board infrastructure for EMG measurement, the MyoWare [5], [21], the Arduino board [3], [22], [23], and the BITalino [24] have often been used. Heywood et al [21] used MyoWare board and incorporated an Analog Device operational amplifier. There was a cable connection between board and electrode. All analyses were performed using LabVIEW on a PC. They reported correlation coefficients of 0.65–0.99, indicating moderate-to-high relative agreement with those of commercial system. Tecchino et al. [23] built a low-cost EMG system with an Arduino Uno ethernet board and an integrated signal conditioning block for biological signals. This system saved data automatically in a secure digital card, and postprocess analysis was done by MATLAB on a PC. During maximal voluntary contraction, an excellent relative agreement was observed (correlation coefficient = 0.30–0.98). Donisi et al. [24] used BITalino sensor, a purpose-built sensor, to measure EMG signals. Several cable codes were required to connection between electrode and main platform. Data were analyzed using MATLAB on a PC. Batista et al. [25] provided evidence validating BITalino, and the computation of the linear envelope resulted in a small difference and high correlation between the BITalino and reference device. Although the price for these specialized board infrastructure for EMG measurement is approximately \$38 (approximatel ¥5,200) to ¥50,000 [4], no studies that employed wireless connectivity and post-processing on the smartphone were found.

III. METHODS

A. SYSTEM DESCRIPTION

The proposed EMG system comprises of two parts, a sensor module and an Android-based smartphone (Fig. 1a). The sensor module captures analog EMG signals via surface electrodes placed on the skin over the muscles of interest. After amplification and filtering, the analog signals are transmitted to an MCU that digitizes the EMG signals using an analog-to-digital (A–D) converter for sampling. A universal serial bus (USB) on-the-go interface is implemented in the MCU to communicate with an Android-based smartphone for data control and transmission. An Android application was also developed to process EMG signals and enable user interaction.

1) HARDWARE

The hardware comprises an electrode, instrumentation amplifier, operational amplifier, a Bluetooth module, lithium polymer rechargeable battery, and small commercially available container (SW-53S, Takachi, Saitama, Japan) with dimensions 35 × 11 × 53 mm (w × h × d) and weight 19.0 g. Furthermore, commercially available stainless-steel balls of diameter 5 mm (SZ54, Hikari, Osaka, Japan) were used as electrodes at distances of 30 mm. The EMG signals obtained from the differential electrodes are amplified using a differential amplifier circuit consisting of an instrumentation amplifier (AD8237ARMZ-R7, Analog Devices, Tokyo, Japan). Differential noise was removed, and the gain of the instrumentation amplifier was set to 20 dB to eliminate the in-phase components of the positive and negative electrodes. The input impedance and common-mode rejection ratio (CMRR) of the differential amplifier were 100 MΩ and 120 dB, respectively. Active electrodes ideally present a significantly high input impedance and low output impedance, resulting in the acquisition systems being considerably robust to power line interference and high electrode-skin impedance values [26]. In this study, the sampling frequency was set to the highest frequency at which no missing data would occur on the smartphone. The EMG signals were bandpass filtered (5–500 Hz) with a sampling frequency of 2,500 Hz, in accordance with the recommendations of the Surface EMG for Non-Invasive Assessment of Muscles (SENIAM) consortium project [27]. The amplitude-frequency characteristic of the primary amplifier acting as a bandpass filter led to the gain being reduced to 20 dB (1/10) outside the 5–500 Hz band. Consequently, two operational amplifiers (LMV358IDR, Texas Instruments, Texas, USA) were placed in series, one as a low-pass and the other as a high-pass filter with threshold frequencies of 500 Hz and 5 Hz, respectively. Their maximum gain was 1,000 (60 dB), which was calculated by multiplying the gains of the instrumentation and operational

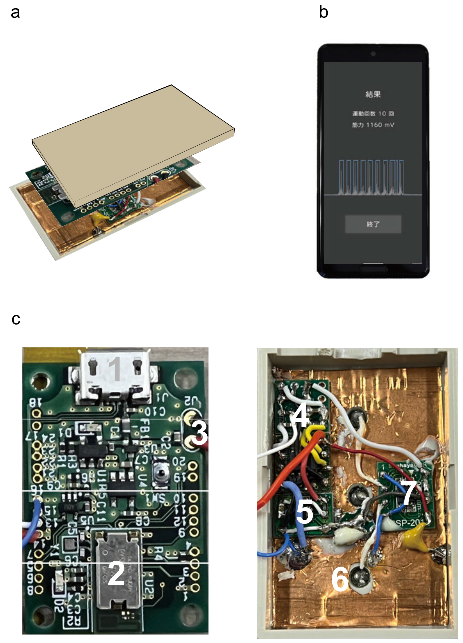

FIGURE 1. **Overall design of the proposed EMG system. (a) Magnified view of the sensor. (b) Smartphone screenshot showing the automatic detection of peak amplitude of EMG signal amplitude during muscle contraction. (c) Image of the circuit board in the sensor showing each functional module: (1) universal serial bus terminal for charging, (2) Bluetooth low-energy module including an advanced reduced instruction set computer (RISC) Machine (ARM) processor and analog/digital converter, (3) connections to the lithium polymer battery, (4) filter implemented by two operational amplifiers, (5) reference voltage generation, (6) electrode, and (7) instrumentation amplifier.**

amplifiers as 10 (20 dB) × 100 (40 dB). An A–D converter and wireless control were used in a Bluetooth low-energy module (EYSKJNZWB, Taiyo Yuden, Tokyo, Japan) containing a 12-bit ARM processor as the central processing unit along with a Bluetooth 5.0 compliant chip and antenna manufactured by Nordic. The analog EMG signals were digitized with 12-bit accuracy and transmitted to a smartphone (Pixel 4a, Google, Mountain View, CA, United States of America) via Bluetooth serial communication. Owing to the wide range of the EMG signal amplitude, a high dynamic range and variable gain are required for the 12-bit A–D converter based on the recommendations of the SENIAM [28].The dynamic range of the A–D converter was 3 V, and its amplification was varied for the EMG signal to reach maximum value. The system was powered by a 110 mAh lithium polymer battery (DTP401525, Data Power Technology, Tokyo, Japan).

2) SOFTWARE

An Android application was developed to visualize and acquire the EMG signals. The application detected the baseline, active, and relaxed phases during muscle contraction bouts. The baseline phase was defined as a resting state for 2–3 s long when the subjects were instructed to fully relax their arm, and the active phase as the time when the EMG signals exceeded four standard deviations (SDs) of the baseline value to just before the relax phase. The relax phase was defined as the time when the EMG signals were less than the baseline value to just before the beginning of the next active phase. The application displayed real-time dynamic charts of the EMG signals during voluntary contraction and automatically detected the amplitude of the EMG signals based on phase classification (Fig. 1b). The EMG data were stored in the internal memory of the smartphone.

IV. EXPERIMENT

A. SUBJECTS

Six healthy male subjects (age: 25.3 ± 9.8 years; height: 170.3 ± 5.6 cm; weight: 69.7 ± 10.8 kg) with no history of neurological disorders participated in this study. Since men fatigue more easily than women during elbow flexion [9], which is the target joint in this study, only male subjects were included. The participants provided informed consent, and all the experimental procedures were approved by the local ethics committee (2022–22). The study was conducted in accordance with the tenets of the Declaration of Helsinki.

B. FORCE AND EMG RECORDINGS

Each subject was seated upright in a neutral position on a chair with the left upper arm and forearm in the vertical and horizontal positions, respectively, and the elbow joint flexed at 90° (Fig. 2). The elbow was supported by a pad to restrain arm movements. The forearm was tightly fixed with a Velcro strap attached to a dynamometer (Biodex System 4, Biodex, New York, USA). The upper arm and forearm were aligned with the principal axes of the dynamometer such that elbow flexion moment primarily produced a force in the upward direction. Before sustained contraction to induce fatigue, the subject performed maximum voluntary contraction (MVC) using the elbow flexor muscles by gradually increasing force from rest to maximum over 3 s, sustaining maximum force for 2–3 s, and then decreasing the force back to baseline.

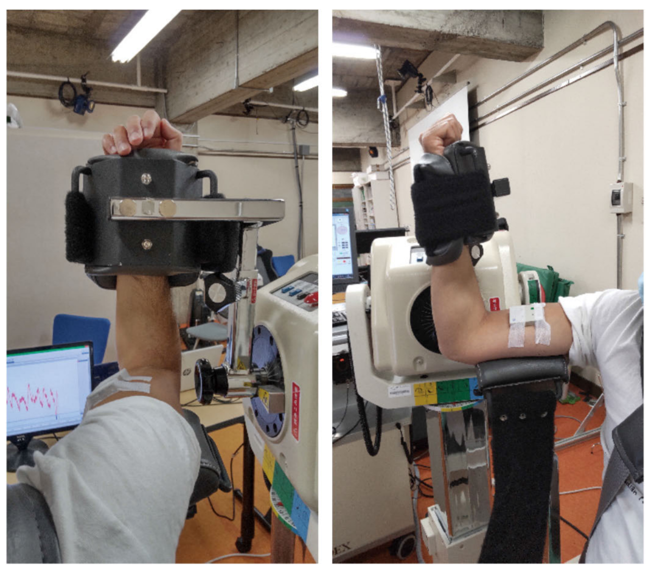

FIGURE 2. **Overview of experimental setup. The subject was seated with arms slightly abducted from the trunk, elbow flexed at 90°, and forearm in a neutral position. The electrode was placed over the biceps brachii muscle.**

The MVCs were repeated three times, and the subject was verbally encouraged to produce maximal effort. A 60 s rest period was incorporated between trials. The greatest force achieved by the subject was taken as the MVC force and used as the reference to calculate the 50% target level for sustained contraction.

Sustained isometric elbow flexion was performed at a target value of 50% of the MVC force. This protocol successfully induced muscle fatigue in a previous study [30]. The subject was required to match the target torque as displayed on the monitor and verbally encouraged to sustain the torque for as long as possible. The sustained contraction was terminated when the force declined by 10% of the target value for longer than ~0.1 s despite verbal encouragement to maintain the task. The force was recorded using an A–D converter (PowerLab 16/35, AD Instruments, Sydney, Australia) at a sampling frequency of 2,000 Hz.

Two measurements were separately obtained using the proposed and commercial EMG systems for research purposes (Trigno, Delsys, Boston, MA, United States of America). The order of measurement was randomized with a rest interval of at least 2 h between each session. After carefully cleaning the skin with alcohol, an electrode was placed on the skin over the short head of the biceps brachii according to the guidelines of surface EMG [27]. The placement of the electrode was marked with a permanent marker to ensure that the electrode was completely placed at the same spot during each measurement. The EMG signals were measured using standard Trigno wireless sensors (10 mm interelectrode distance, CMRR > 80 dB @60 Hz, gain of 100, transmission range of 40 m, durable battery lifetime of 4–8 h, 16-bit resolution, 500μs inter-sensor latency) and amplified and bandpass filtered (20–450 Hz) using an A–D converter (PowerLab 16/35, ADInstruments) at a sampling frequency of 2,000 Hz. The price of a Trigno EMG system package (16 channels) is ¥4,751,250 (Price at time of purchase). Table 1 shows a comparison between the key features of our proposed and the referenced systems.

TABLE 1. Feature comparison of proposed and referenced systems.

Feature	Our proposed system	Referenced system
Model/company		Trigno/Delsys
Frequency (Hz)	2,500	2,000
Resolution (bits)	12	16
Bandwidth (Hz)	5–500	20–450
Interelectrode distance (mm)	30	10
Gain (dB)	1000	100
CMRR (dB)	120	80
Impedance (MΩ)	100	10,000
Battery (mAh)	110	

C. DATA ANALYSIS

The EMG signals acquired via both the systems were analyzed using an in-house MATLAB scrip (R2017a, Mathworks, Natick, MA, USA). The EMG signals were segmented into 10 s epochs. For each epoch, the root mean square (RMS) and mean power frequency (MPF) of the EMG signals (EMG_{RMS} and EMG_{MPF}, respectively) were calculated based on the Fourier transform. EMG_{RMS} was calculated to examine muscle fatigue-induced changes in the amplitude of muscle activity, defined as follows

$$EMG_{RMS} = \sqrt{\frac{\sum_{i=1}^{n} |rawData_i|}{n}}, \tag{1}$$

where i is the order number of the dealing sample point, raw data is the value of the ith sample point, and n is the total number of data points.

To examine the muscle fatigue-induced changes in the characteristic spectral frequency of EMG signals, EMG_{MPF}, calculated as the frequency centroid of the spectrum, is defined as follows:

$$EMG_{MPF} = \frac{\int_0^\infty S(f) \cdot f \cdot d(f)}{\int_0^\infty S(f) \cdot fd}, \tag{2}$$

where f is the frequency, $S(f)$ is the power at frequency f, and $d(f)$ is the frequency resolution.

EMG_{RMS} and EMG_{MPF} were calculated for 10 s in steps of 1 s [31]. The initial and final values for EMG_{RMS} and EMG_{MPF} were defined as the beginning and ending values of the sustained isometric contraction, respectively. The relative changes in EMG_{RMS} and EMG_{MPF} were also defined as the final value normalized by the initial value. The signal-to-noise ratio (SNR) was calculated as the value of the raw EMG signal during sustained muscle contraction relative to the value of the baseline noise. Battery lifetime was tested for continuous EMG signal measurement and transmission to a smartphone via Bluetooth.

D. STATISTICS

All data are presented as mean ± standard deviation. The difference in the SNR of the EMG signal and the normalized EMG_{RMS} and EMG_{MPF} between the two EMG systems was evaluated using paired t-tests. EMG_{RMS} and EMG_{MPF} were analyzed using repeated measures two-way analysis of the variance (time × EMG system). Post-hoc tests (Bonferroni pairwise comparisons) were used to investigate the eventual differences between the two EMG systems. The relative agreement of the given variables as assessed by both the EMG systems was tested using the intra-class correlation coefficient (ICC)(2,1). Estimates of correlation were interpreted as excellent (0.75–1), modest (0.4–0.74), or poor (0–0.39) [32]. The significance level was set at $P < 0.05$.

V. RESULTS

The mean time to complete a sustained isometric contraction for all the subjects was 83.5 ± 12.4 s (range: 68–96 s) for the proposed EMG system and 87.2 ± 23.1 s (range: 52–108 s) for the commercial EMG system (Fig. 3). The SNR was significantly lower in the proposed EMG system (77.4 ± 6.8 dB) than in commercial EMG systems

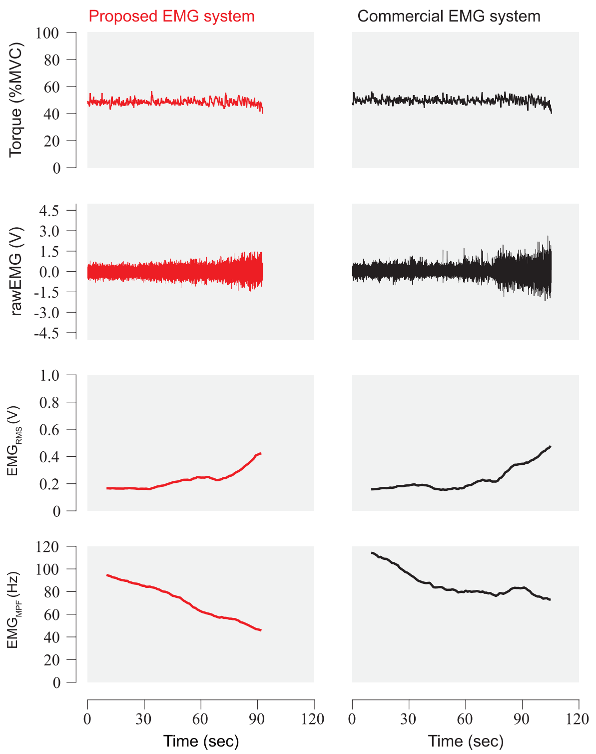

FIGURE 3. Typical example (n = 1) of progressive changes in torque, raw EMG, EMG_{RMS}, and EMG_{MPF} during sustained isometric contraction of the elbow flexor muscles at 50% of MVC till the endurance limit was reached. Two measurements were performed: the red indicates our proposed EMG system and black indicates the commercial EMG system. EMG_{RMS} and EMG_{MPF} were calculated over a 10 s window along the signal in 1 s steps.

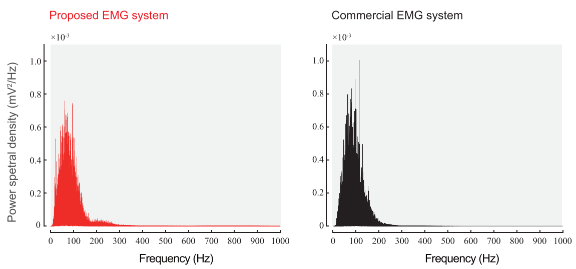

FIGURE 4. Typical example (n = 1) of the power spectral density of the surface EMG signals during sustained isometric contraction of the elbow flexor muscles at 50% of MVC, till the endurance limit was reached. Two measurements were performed: the red indicates our proposed EMG system and black indicates the commercial EMG system.

(116.5 ± 18.8 dB) (P < 0.001). No significant difference in power spectral density was observed between the two EMG systems (Fig. 4). The interaction between time and EMG systems was not significant for EMG_{RMS} and EMG_{MPF}. However, it was significant for EMG_{RMS} ($P = 0.002$) and EMG_{MPF} ($P < 0.001$), suggesting that regardless of the EMG system, EMG_{RMS} was significantly higher in the final value than in the initial value, while EMG_{MPF} exhibited the opposite trend (Fig. 5). The effect of the EMG system was significant for EMG_{MPF} ($P = 0.002$), suggesting that, regardless of time, EMG_{MPF} showed significantly lower values in the proposed EMG system. Normalized EMG_{RMS} demonstrated an excellent agreement (ICC = 0.933, $P < 0.001$) between

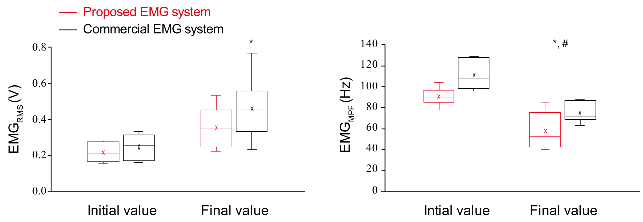

FIGURE 5. Comparisons for EMG_{RMS} (left) and EMG_{MPF} (right) of the raw EMG in the initial and final values of sustained isometric contraction between the two EMG systems. The cross represents the average value. Data from the proposed EMG system are denoted with red boxplots while those from the commercial EMG system are denoted with black boxplots; * indicates the main effect of time, P < 0.05, and # indicates the main effect of the EMG systems, P < 0.05.

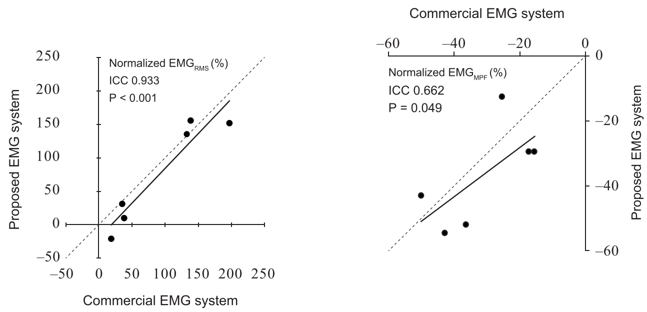

FIGURE 6. Normalized, defined as the final value normalized by the initial value, EMG_{RMS} (left), and EMG_{MPF} (right) of the raw EMG measured by the proposed and commercial EMG systems. The dashed line is the line of identity, and the solid line is the line of best fit.

TABLE 2. List and cost of hardware components.

List	Product name, manufacturing company (location)	Cost (Japanese Yen)
Stainless-steel ball	SZ54, Hikari (Osaka, Japan)	23
Lithium-ion polymer battery	DTP401525, Data Power Technology (Tokyo, Japan)	879
Enclosure of sensor module	SW-53S, Takachi Yuden (Tokyo, Japan)	131
Bluetooth module	EYSKJNZWB, Taiyo Yuden (Tokyo, Japan)	3,190
Instrumentation Amplifier	AD8237ARMZ-R7, Analog Devices (Tokyo, Japan)	440
Operational Amplifier	LMV358IDR, Texas Instruments (Texas, USA)	204
Board development	Towa Electronics (Tokyo, Japan)	3,619
	Subtotal	8,486
Smartphone	Pixel 4a, Google (California, USA)	32,890
	Total	41,376

the two EMG systems, while normalized EMG_{MPF} indicated modest agreement (ICC = 0.662, $P = 0.049$) (Fig. 6). No significant difference was observed in the normalized EMG_{RMS} and EMG_{MPF} between the proposed and commercial EMG systems. Our system operated for over 8 hours on a 110 mAh battery. Lastly, Table 2 lists all the hardware components and their costs. The total cost of hardware components was ¥41,376 per unit, and the price of the smartphone was 79.5% of the total cost.

VI. DISCUSSION

This is the first study to develop a versatile EMG for consumer use and to validate EMG signals during fatiguing contraction. As raw EMG signals are significantly sensitive

TABLE 3. Key characteristics comparison of our proposed system with commercial devices.

Model name, Company	Electrode type, Weight Size (mm)	Resolution (bit)	Sampling Frequency (Hz)	Signal Bandwidth (Hz)	SNR (dB)	Range (m)	Battery life (hours)	Cable connection	Data store/ transmit	Postprocess apparatus	# of channels	Cost (JPY¥)
Our proposed system	Dry, 19 g 35 × 11 × 53	12	2500	5–500	77[1)]	7	> 8	No	Bluetooth	Smartphone	1	¥8,486 (sensor)
Trigno, Delsys	Dry, < 15 g 27 × 15 × 37	16	2000	20–450	117[1)]	40	4–8	No	Bluetooth	PC	16	¥8,800,000 (sensor, base station, software, power supply)
BiostampRC, MC10	Wet, 7 g 34 × 4.5 × 66	16	500, 1000	20–450	8.2[2)]	5	< 28	No	Bluetooth	Tablet	1	¥680,000
mDurance, mDurance Solution	Wet, 31 g 32 × 12 × 65 (Bridge Amplifier)	24	1024	8000	21–23[3)]	10	4.5	Yes (Bridge Amplifier to electrode)	SD card, Bluetooth	Smartphone	4	EUR €4,500 (¥650,000)
MyoWare, Advancer Technologies	Wet, 28 g 52 × 21	10	1100	10–400	11-18[3)]	NA	NA	Yes (Sensor to electrode)	Wired	PC	1	USD $38 [6)] (¥5,200) (sensor)
EMG One, EMG One	Wet, 350 g 15 × 5 × 19 (Adaptor)	16	1000	70–380	NA	NA	NA	Yes (adaptor to electrode)	Wired	Smartphone, Tablet	1	USD $357 (¥49,000) (sensor)
Athos, Athos,	Textile, 22 g	NA	1000	> 500	NA	60	10	No	Bluetooth	Smartphone	1	USD $348 [6)] (¥47,500)
Duelite, OT Bioelettronica	Wet, 12 g 21 × 11 (acquisition probe)	16	2048	10–500	18–23[4)]	< 9	13	Yes (acquisition probe to electrode)	Bluetooth with USB dongle	PC, Tablet	2	¥232,000 (sensor)
mTrigger Biofeedback, mTrigger	Wet, 113 g 102 × 61 × 28 (Biofeedback unit)	NA	NA	NA	NA	NA	NA	Yes (post-process apparatus to electrode)	Wired	Smartphone, Tablet	2	USD $449 (¥61,500)
PicoEMG, Cometa	Wet, 7 g 41 × 10 × 16	16	2000	1–10	4.2	10	< 12	No	Wifi	PC	2	¥2,640,000 (sensor and software)
LE250, Biometric	Dry, 17 g 24 × 14 × 42	13	500, 1000, 2000	5–495	21–25[3)]	30	< 8	No	Bluetooth with USB dongle	PC, Tablet	1	¥984,500 (sensor, software, dongle, power supply)
EMG sensor, Plux Biosignal	Wet, 45 g 85 × 10 × 54 (Hub)	8, 16	< 4000	25–500	NA	10	> 10	Yes (wireless hub to electrode)	Bluetooth	Smartphone, Tablet, PC	4	¥217,800 (sensor, hub)
BITalino, Plux	Wet, 2 g 100 × 65 × 6	6, 10	1, 10, 100, 1000	25–480	55.7[5)]	10	8	Yes (core board to electrode)	Bluetooth	Smartphone, Tablet, PC	1	¥50,000 (sensor, hub)

The list of commercial devices was added based on previous studies [4,34].
EUR; euro, JPY; japanese yen, NA; not applicable, PC; personal computer, SD card; secure digital card, SNR; signal-to-noise ratio, USD; united states dollar, USB; universal serial bus. SNR data from [1)] this study; [2)] Jang et al. [35]; [3)] González-Mendoza et al. [36]; [4)] Zhao et al. [20]; [5)] Guerreiro et al. [37]. Price from [6)] Manzur-Valdivia and Alvarez-Ruf [4]; other prices are based on inquiries with the manufacturer (Dec 2022) and website.

to noise from the acquisition device and measurement environment, investigating the EMG signals obtained by the custom-made product to verify their similarity to those by a reference product is important. The obtained results demonstrate that the relative agreement between the proposed and commercial EMG systems is significant for EMG_{RMS} (ICC = 0.933, $P < 0.001$) and EMG_{MPF} (ICC = 0.662, $P = 0.049$). Few studies have validated the use of custom-made, low-cost EMG systems [6], [15], [16]. For example, Yang et al. [16] implemented a low-cost surface EMG acquisition system combined with an MCU and a Wi-Fi module. They calculated the mean absolute value of raw EMG during comfortable walking. The ICC between custom-mode and commercial device was 0.86 for the vastus lateralis and 0.81 for the biceps femoris. To our best knowledge, no studies have compared custom-made and commercial EMG systems during fatiguing contraction. According to one of the EMG device-manufacturing companies, 1.2 dB is an acceptable SNR [33]. A higher SNR indicates a better signal quality. The SNR of the proposed EMG system was 77.4 ± 6.8 dB, while that of the commercial EMG system was 116.5 ± 18.8 dB. Sensor location and skin preparation, which influence SNR, are completely identical for our proposed and commercial EMG systems. Consequently, the influence of physical factors for the SNR difference between the two EMG systems is eliminated. Hence, the hardware components may be involved. The CMRR determines the amount of common noise that can be removed from the signal. Additionally, it has a direct impact and is approximately proportional to the SNR [11]. Nevertheless, the SNR of our proposed system is significantly lower than that of commercial systems, possibly due to differences in input impedance. To prevent attenuation and distortion of the detected signal due to the effects of input loading, the input impedance of the differential amplifier should be as large as possible [33]. Fu et al. [11] reported that an SNR between 37.0–46.2 dB using a low-cost EMG amplifier with discrete operational amplifiers. In addition, Prakash et al. [13] compared the SNR between self-made EMG sensors (18.9 dB) and commercially available MyoWare sensor (4.8 dB). Therefore, this proves that the proposed EMG system not only detects fatigue accurately, but also achieves adequate noise immunity.

Accordingly, the proposed EMG system was successfully fabricated using easily available and low-cost components; moreover, its circuit board was developed in-house, and the system is controlled with a smartphone. The list of commercial devices in Table 3 was refers based on the previous studies [4], [34]. Only MyoWare and our proposed system cost less than ¥10,000. Although limited studies mention the market prices of the packaged EMG systems, some commercial products are significantly expensive, with prices ranging from approximately €15,000 to €20,000 (approximately ¥2,200,000 to ¥2,942,000) [3] or up to $20,000 (approximately ¥2,750,000) [4]. In contrast, several EMG sensors are sold in the market. del Toro et al. [3] reported the cost of their system in the range of €100–€150 (approximately ¥14,700–¥22,000), which is not significantly different from the price of the proposed EMG system. However, EMG sensors face several challenges. One problem is that postprocessing must be performed by the user, which is difficult. Additionally, the cables extending from the sensors are wired to the amplifier, which are detrimental to the EMG measurements. In contrast, the proposed EMG system is completely wireless and accompanied by a smartphone application that handles the postprocessing, making it superior to the EMG systems available in the market. Consequently, our proposed system has better usability than other commercial devices considering highly versatile due to wireless and smartphone accessibility.

Previous studies have indicated that relative anatomical alignment and movement of the muscle fibers, location of the innervation zone, and crosstalk between muscles can influence EMG signals [38]. Different electrode positions and interelectrode distances can also affect EMG signals [39]. In this study, both EMG systems were derived the signal from the exact same anatomical location; therefore, the observed differences cannot be attributed to their structural factors. The observed differences are primarily attributed to the differences in the electronic components. One potential possibility is the difference in inter-electrode distances. The 30 mm inter-electrode distance of the proposed EMG system is generally recommended for bipolar surface EMG [27] since it is larger than that of the Trigno sensor. Theoretical, experimental, and simulation investigations have established that the frequency content of a surface EMG signal decreases with increasing inter-electrode distance because of increased detection volume and reduced spatial filtering of the signal [40], [41], [42]. Furthermore, the commercial EMG system has a larger gain (100 vs. 20 dB) and higher quality A–D converter (16 vs. 12 bits), thereby enhancing signal amplification when EMG activity is relatively low. This suggests that to investigate the EMG signal during submaximal contractions, attention must be paid when using low-cost EMG systems.

Furthermore, EMG_{RMS} progressively increases and EMG_{MPF} decreases during sustained submaximal isometric contraction, where the force level is constant until exhaustion. When the muscles are fatigued, the nerve conduction velocity decreases, and the low-frequency component of EMG signals increases, increasing the amplitude of EMG signals by allowing more energy to reach the electrodes owing to the low-pass filter effect [43]. The proposed EMG system developed in this study succeeded in reproducing these changes in EMG signals associated with muscle fatigue in accordance with the aforementioned theory.

The proposed EMG system opens the possibility of continuous muscle fatigue monitoring during everyday life of the user. One advantage of our system is that it sends the data of the patien's or athlete to the physiotherapists or clinicians in real time. This system will reduce treatment time and costs as it will allow the patient or athlete to implement rehabilitation exercises remotely and reduce the number of routine visits to the clinic. Our findings indicate an excellent agreement between our proposed and commercial EMG systems during fatiguing contraction, although the various limitations to this study must be noted. Firstly, as we aimed to produce a low-cost EMG sensor using commercially available inexpensive components, a single channel and low battery life are observed. In the future, multi-channel systems and extended battery life will be required for daily use. Secondly, to design a controlled experimental environment, the results presented here were limited to the biceps brachii muscle and isometric contraction in a relatively small group of primarily younger healthy subjects, without considering changes associated with aging or pathological conditions. The features of the surface EMG signal and changes with fatigue may differ across muscles, with subject anatomy and subcutaneous fat thickness. In particular, the tissue between the electrode and muscle fibers has a spatial low-pass filtering effect on the EMG signal detected at the skin surface, with decreases both the amplitude and frequency content of the signal [44]. Despite these limitations, establishing signal properties under isometric conditions in healthy subjects is a critical first step before progressing to a wider range of conditions.

VII. CONCLUSION

The novelty of the present study is that we have developed a consumer-friendly EMG system that is easy to use (entirely wireless, smartphone-controlled, and battery-powered) and highly accurate, and low-cost (less than $100 (approximately ¥10,000 yen)). Experimental results indicate that the proposed EMG system, which is highly versatile owing to its wireless design and smartphone accessibility, can accurately detect changes in EMG amplitude and frequency during fatiguing isometric contractions. This EMG system can monitor EMG signals associated with muscle fatigue during daily activities of the user. This study enables the possibility of further evaluating stiff shoulders, back pain, and swallowing difficulties. Additionally, we aim to support the early prediction of diabetes among healthy individuals, which has been difficult to evaluate with conventional EMG devices.

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