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Dynamic monitoring of forearm muscles using one-dimensional sonomyography system

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Abstract—We introduce a method, known as one-dimensional sonomyography (1-D SMG), that uses A-mode ultrasound signals to detect dynamic thickness changes in skeletal muscle during contraction. We custom-designed a 1-D SMG system to collect synchronized A-mode ultrasound, joint angle, and surface electromyography (EMG) signals of forearm muscles during wrist extension. We extracted the 1-D SMG signal from the ultrasound signal by automatically tracking the corresponding echoes, which we then used to calculate muscle thickness changes. We tested the right forearm muscles of nine nondisabled young subjects while they performed wrist extensions at 15.0, 22.5, and 30.0 cycles/min and their largest wrist extension angle ranged from 80° to 90°. We found that the muscle deformation and EMG root mean square signals correlated linearly with wrist extension angle. The ratio of deformation to wrist angle was significantly different among the subjects ($p < 0.001$) but not among the trials of different extension rates for each subject ($p = 0.9$). The results demonstrate that 1-D SMG can be reliably performed and that it has the potential for skeletal muscle assessment and prosthesis control.

Key words: electromyography, EMG, forearm muscles, mechanomyography, MMG, rehabilitation, skeletal muscles, SMG, sonomyography, ultrasound, wrist extension.

INTRODUCTION

Electromyography (EMG), an electrical signal collected by electrodes during muscle contractions, represents the bioelectrical properties of skeletal muscles and demonstrates the physiological processes of muscle contraction. It has been widely used for evaluation of muscle function in the areas of biomechanics and kinesiology [1–2], muscle pathology [3], muscle fatigue [4], and prosthetic device control [5]. The root mean square (rms) magnitude and median frequency are commonly used to describe the time-domain and frequency-domain information of the EMG signal, respectively [6].

EMG is a complex signal; it is the summation of individual motor unit (MU) action potential trains generated by irregular discharges of active MUs during muscle activation. It can be influenced by many factors, e.g., muscle cross talk [7] and interelectrode distance [8]. During the past decade, many efforts have been directed at developing different algorithms to process EMG signals, including classification of EMG using artificial network [9], fuzzy logic [10], and pattern recognition (multichannel EMG

Abbreviations: 1-D = one-dimensional, ANOVA = analysis of variance, EMG = electromyography, ICC = intraclass correlation coefficient, MK = myokinemetric, MMG = mechanomyography, MU = motor unit, SD = standard deviation, SMG = sonomyography, rms = root mean square, UMME = Ultrasound Measurement of Motion and Elasticity.

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[11]) and decomposition of EMG signals with the Bayesian method [12]. However, in addition to the complexity of the required signal processing methods, use of EMG for noninvasively measuring deep muscles is difficult because the deep muscle EMG signal may be more attenuated and/or mixed with the superficial muscle EMG signal by the time it reaches the skin surface.

Researchers have been searching for alternative signals that can better assess muscle function, including mechanomyography (MMG), electroencephalography [13–14], myokinemetric (MK) signals [15], and magnetic resonance imaging [16]. For example, MMG is the sound generated by a muscle during its contraction and is used as a measure of mechanical muscle changes during contraction [17]. Recently, it has been widely analyzed along with EMG for different purposes [18–20], such as control of a prosthesis with 2 degrees of freedom [21]. However, MMG can be affected by many factors, such as muscle temperature [22], skinfold thickness [23], and external mechanical noise [24]. These factors, together with challenges in sensor attachment and low-frequency noise elimination, can affect the stability and reliability of the MMG signal, thus limiting its application in fatigue assessment and prosthesis control. The MK signal represents the dimensional changes of muscle while it bulges during contraction and is detected with a Hall sensor [15]. Sensor attachment is also a challenge when collecting the MK signal.

Because it has the advantages of being stable, easy to use, nonionizing, and capable of recording activities from deep muscles without cross talk from adjacent muscles [25], ultrasonography has been used to detect muscle thickness changes [26–27], pennation angle [28–29], cross-sectional area [30–31], and muscle fascicle length [32–34] both in static and quasi-static conditions during the past decades. Since skeletal muscle architecture is closely correlated with its function [35], ultrasound parameters have been widely employed to characterize muscle activity [36–38], and the relationship between EMG and muscle architecture changes detected with ultrasound has been reported [39–40].

In a previous study, Zheng et al. used sonomyography (SMG) with B-mode ultrasound images to describe real-time muscle thickness changes during contraction [41]. A system was developed to record and analyze ultrasound images, force, joint angle, and surface EMG simultaneously. The system has been successfully used for the analysis of muscle fatigue, and the investigators found that muscle thickness increased during the fatigue process [42]. The correlation between EMG and SMG of muscles during isometric contraction has also been investigated [43].

Although the two-dimensional SMG signal from ultrasound images is capable of detecting continuous muscle thickness changes, A-mode ultrasound, with a more portable and compact transducer, should be a less expensive and more practical alternative for detecting muscle thickness changes during contraction. In this study, we used a system equipped with an A-mode ultrasound transducer (one-dimensional SMG [1-D SMG]) to detect thickness changes in forearm muscles. The relationships between the surface EMG signal and wrist angle and between the 1-D SMG signal and wrist angle were quantitatively studied. The results were used to assess the potential of 1-D SMG signals as a noninvasive method for detecting skeletal muscle activity in vivo.

MATERIALS AND METHODS

Subjects

Nine nondisabled subjects (seven male and two female), aged 24 to 35 years, were recruited for the experiment. All subjects had no known neuromuscular disorders. Human subject ethical approval was obtained from the relevant committee at The Hong Kong Polytechnic University, and informed consent was obtained from all subjects before the experiment.

Experimental Setup and System Calibration

As shown in **Figure 1**, an ultrasound pulser/receiver (GE Panametrics, model 5052 UA, General Electric Company; Fairfield, Connecticut) was used to drive a 10 MHz single-element ultrasound transducer (GE Panametrics, model V129) and amplify the received signals. We digitized the A-mode ultrasound signal using a high-speed analog-to-digital converter card at a sampling rate of 100 MHz (Gage model CS82G, Gage Applied Technologies, Inc; Lockport, Illinois). The angle signal during wrist extension was measured with an electronic goniometer (model XM110, Penny & Giles Biometrics, Ltd; Gwent, United Kingdom). We captured the surface EMG signal from the EMG bipolar Ag-AgCl electrodes (Axon Systems, Inc; Hauppauge, New York) and preamplified it with a gain of 100 using a circuit located near the electrodes. The signal was further amplified by another factor of 10 by the custom-designed EMG amplifier and filtered

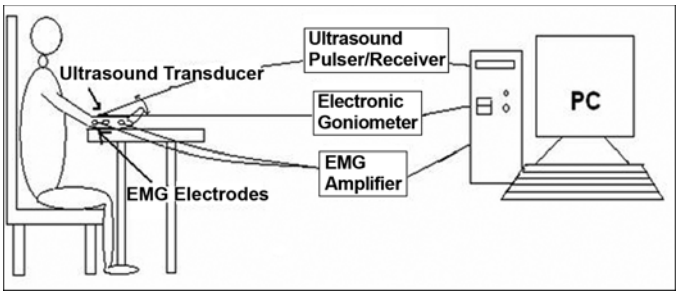

Figure 1.
Data collection system. EMG = electromyography, PC = personal computer.

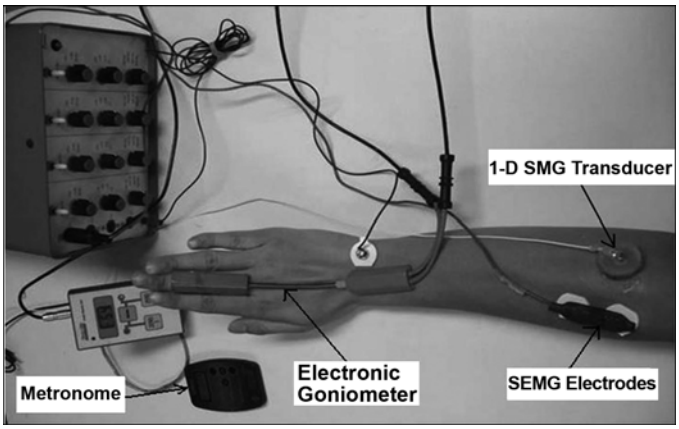

Figure 2.
Placement of one-dimensional sonomyography (1-D SMG) transducer, surface electromyography (SEMG) electrodes, and electronic goniometer on forearm. Ultrasound coupling gel was applied between transducer and skin.

by a 10–300 Hz band-pass analog filter within the EMG amplifier. The surface EMG and wrist angle signals were digitized with a 12-bit data acquisition card (model NI-DAQ 6024E, National Instruments Corp; Austin, Texas) at a sampling rate of 4 KHz. The A-mode ultrasound signal was saved frame by frame along with the surface EMG and wrist angle signals for subsequent analysis.

The 10 MHz single-element ultrasound transducer (diameter 5 mm) was selected because this frequency could satisfy the required resolution and scanning depth in this study. The transducer was inserted into a custom-made holder (diameter 26 mm) that was made of silicone gel so we could attach it to the skin (**Figure 2**). We used a portable B-mode ultrasound scanner (180 Plus, SonoSite Inc; Bothell, Washington) to identify the location of the extensor carpi radialis. Before data collection, each subject was asked to perform several wrist extensions through which the most prominent bulge in the extensor carpi radialis belly was identified. The single-element ultrasound transducer was then positioned on the skin above the identified bulge and fixed by double-sided adhesive tape. We applied ultrasound gel between the transducer and the skin to aid acoustic coupling. We attached the EMG bipolar Ag-AgCl electrodes to the skin surface of the extensor carpi radialis belly near the ultrasound transducer at a distance of ~1 cm to avoid potential effects from the ultrasound coupling gel [44]. The distance between the two electrodes was 20 mm, and their orientation was parallel to the extensor carpi radialis muscle fibers. An additional reference electrode was placed near the head of ulna. We placed the electronic goniometer in the middle of the posterior hand to measure the wrist angle during wrist extension.

The A-mode ultrasound, surface EMG, and wrist angle signals were collected, stored, and analyzed with the Ultrasound Measurement of Motion and Elasticity (UMME) software (http://www.tups.org/) developed with Microsoft Visual C++ (Microsoft Corp; Redmond, Washington). The time delay among the three data collection systems was calibrated with a method similar to that described by Huang et al. [45]. As the transducer moved cyclically up and down in a water tank, the three signals representing A-mode ultrasound, surface EMG, and wrist angle were collected and stored. The time delay among the three signals was calculated with a cross-correlation algorithm.

Experimental Protocol

The right forearms of the subjects were chosen for testing because they were the dominant ones for all subjects. Each subject was seated in a comfortable chair with his or her forearm resting on the table. After skin preparation with alcohol swabs and several warm-up contractions, each subject was asked to perform wrist extension guided by a metronome (model MT-40, Wittner GmbH & Co; Allgäu, Germany) at three extension rates of 15.0, 22.5, and 30.0 cycles/min. For each extension rate, three repeated tests were performed, with a rest of 3 minutes between two adjacent trials, and three wrist extension cycles were performed in each trial. During the experiment, subjects were given continuous encouragement to try their best to reach their largest wrist extension angle, which for most subjects was between 80° and 90°.

The A-mode ultrasound signals were collected by the UMME software at a sampling rate of 100 MHz and saved frame by frame. The surface EMG sampling was synchronized with the collection of the ultrasound signals. Each frame contained 8,192 sampling points of reflected ultrasound echo data, equivalent to a depth of approximately 6 cm, accompanied by an EMG epoch of 64 ms and a wrist angle value. The frame rate was 14 Hz, and 100 frames of data were recorded for each test.

Data Analysis

We used a cross-correlation algorithm to track the displacement of the upper and lower boundaries of the extensor carpi radialis muscle during wrist extension. The algorithm required a reference signal from an initial frame and would search for the most similar signal to the reference signal to estimate the object position in the updated frame. In this study, we were most interested in the A-mode ultrasound echoes reflected from the fat-muscle and muscle-bone interfaces. When the muscle was contracting, its dimensional changes induced variations in the distance between these two interfaces, which in turn caused the A-mode ultrasound echoes to shift for a certain distance. We selected the A-mode ultrasound signals using two tracking windows, and the correlation tracking algorithm tracked the movement of the selected echoes frame by frame automatically. The width of the tracking window was selected manually and included enough echo features for reliable tracking. This manual selection of the tracking window slightly affected the muscle deformation value, but its effect on the percentage change should be negligible. The distances between the fat-muscle and muscle-bone interfaces were calculated for each frame. The percentage deformation of muscle (D) was defined as

$$D = \frac{(d - d_0)}{d_0} \times 100 \ ,$$

where d_0 was the initial distance between the two echoes and d was the distance during muscle contraction.

The rms amplitude of the surface EMG was calculated for each 64 ms epoch and smoothed using a program written in MATLAB, version 6.5 (The MathWorks, Inc; Natick, Massachusetts). We used wavelet shrinkage as a smoothing operator to reduce the fluctuation in the EMG rms signal [46]. We observed that this method could effectively smooth the signal, while at the same time better preserve the required signal components.

To investigate the reliability of using A-mode ultrasound signals to describe wrist movement, we studied the relationship between 1-D SMG and wrist angle signal using linear regression. We used one-way analysis of variance (ANOVA) to analyze the deformation-angle ratios among trials with different extension rates and subjects. We calculated intraclass correlation coefficients (ICCs) using SPSS (SPSS Inc; Chicago, Illinois) for the repeated trials for each rate. Statistical significance was set at the 5 percent probability level.

RESULTS

Relationship Between Muscle Deformation and Wrist Angle

Figure 3 shows the muscle deformation and wrist extension angle of a typical trial obtained from subject 1. The three cycles indicate the three repeated wrist extension processes in a single trial. The ascending part of each cycle corresponds to muscle contraction, and the descending portion corresponds to muscle relaxation. Using linear regression, we found that muscle deformation correlated very well with wrist extension angle, with a correlation coefficient of $r = 0.91$ (**Figure 4**). The experimental data obtained from the other subjects showed a similar result, and the overall mean r value for the nine subjects was 0.91 ± 0.08 (mean ± standard deviation [SD]).

We also studied the relationship between wrist extension angle and the corresponding muscle deformation using their ratio, i.e., the slope of the linear regression in **Figure 4**. The overall mean value of the deformation-angle ratio was 0.130 %/° ± 0.058 %/° (mean ± SD). One-way

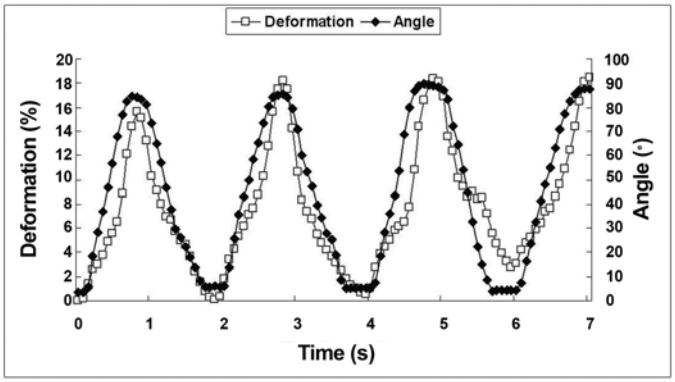

Figure 3.
Muscle deformation and wrist extension angle of typical trial with 3 wrist extension cycles for subject 1.

ANOVA demonstrated that the ratios were significantly different among the subjects ($p < 0.001$) but not among the trials of different extension rates for each subject ($p = 0.9$). **Figure 5** and the **Table** show the summary of the muscle deformation to wrist angle ratios for individual subjects at different extension rates. The ICC for the three repeated trials was 0.87, indicating good repeatability.

Relationship Between Surface EMG and Wrist Angle

Figure 6 shows a typical relationship between the EMG rms and wrist angle based on the results obtained from subject 1. The relationship could also be represented by a linear regression with correlation coefficient $r = 0.87$ (**Figure 7**). The results obtained from the other subjects showed a similar trend. The mean r value of the tests for all the subjects was 0.86 ± 0.07 (mean ± SD). The relationship between the EMG rms signal and wrist angle showed poorer correlation compared with that between the 1-D SMG signal and wrist angle.

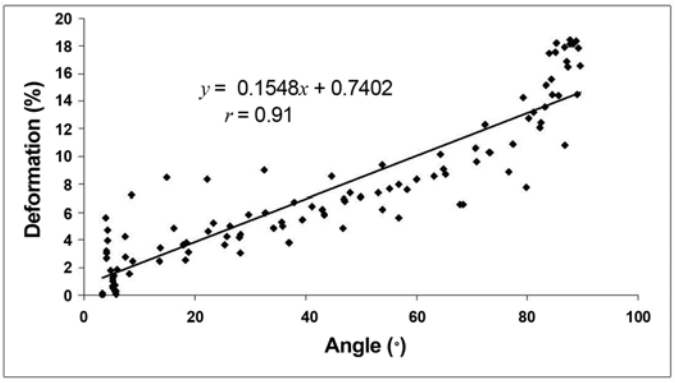

Figure 4.
Relationship between percent muscle deformation and wrist extension angle of typical trial for subject 1. Linear regression was used to represent correlation between two signals.

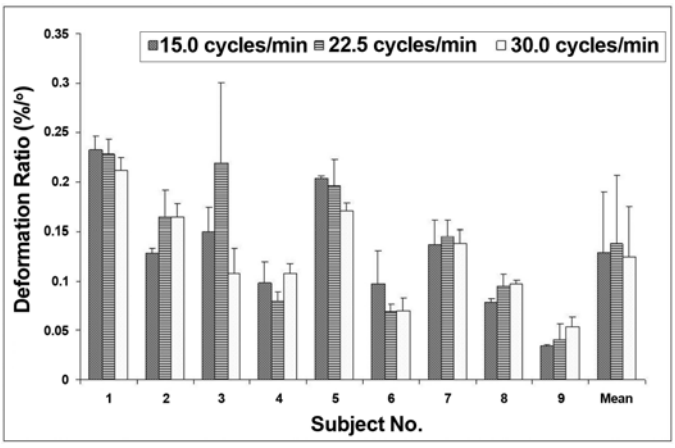

Figure 5.
Ratios between muscle deformation and wrist angle for subjects 1–9 tested at wrist extension rates of 15.0, 22.5, and 30.0 cycles/min. Error bars represent standard deviation of results of 3 trials for each subject at each rate. Last bar (mean) represents overall mean and standard deviation of results of all subjects.

DISCUSSION AND CONCLUSIONS

In this study, we used 1-D SMG with A-mode ultrasound signals and a single-element ultrasound transducer to detect thickness changes in forearm muscles during

Table.
Mean ± standard deviation of ratio between muscle deformation and wrist extension angle for 9 nondisabled subjects. Results were calculated from data obtained at 3 different wrist extension/flexion rates.

Subject No.	Ratio (%/°)
1	0.224 ± 0.011
2	0.152 ± 0.021
3	0.159 ± 0.056
4	0.095 ± 0.014
5	0.190 ± 0.017
6	0.079 ± 0.016
7	0.140 ± 0.004
8	0.090 ± 0.010
9	0.043 ± 0.010

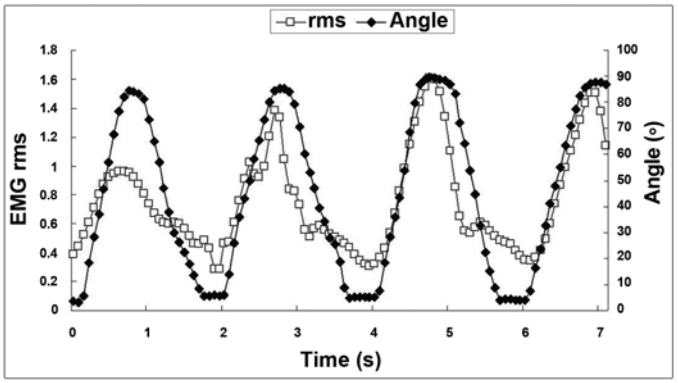

Figure 6.
Smoothed electromyography (EMG) root mean square (rms) and wrist extension angle of typical trial with 3 cycles of wrist extension for subject 1.

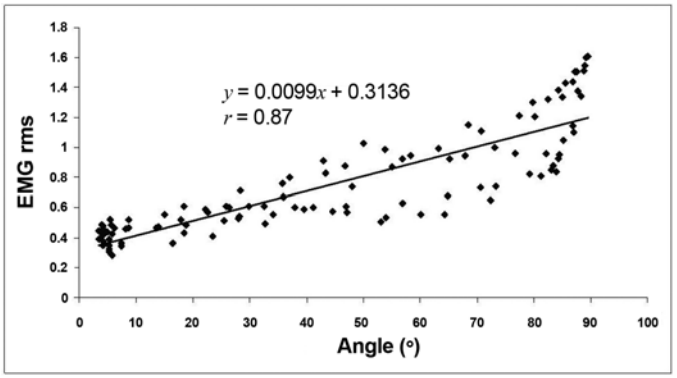

Figure 7.
Relationship between smoothed electromyography (EMG) root mean square (rms) and wrist extension angle of typical trial for subject 1. Linear regression was used to represent relationship between smoothed surface EMG rms and wrist extension angle.

contraction [41]. We found a good linear correlation (r = 0.91 ± 0.08) between the 1-D SMG signal and the corresponding wrist extension angle for the subjects tested (n = 9), demonstrating that the morphological changes of forearm muscles were significantly correlated to the motion of the related joint. We also calculated the deformation-angle ratio to further study the relationship between muscle thickness change and wrist extension angle. This ratio was found to be significantly different among the nine subjects (**Table**). However, for each subject, no significant differences were observed for the ratios obtained at different wrist extension rates. Therefore, a single ratio can describe the relationship between muscle deformation and wrist angle signal for one subject. More experiments on subjects of different ages and sexes are required to better understand the factors that affect the deformation-angle ratio. The surface EMG as a reference signal was collected simultaneously, and the results demonstrate that a significant linear correlation also existed between the EMG rms and wrist angle (r = 0.86 ± 0.07). Since surface EMG has been widely used to assess muscle function and prosthesis control, we expect that 1-D SMG can also be used to provide more information about muscle thickness changes.

Recently, muscle thickness change has been used as a parameter to characterize muscle fatigue [42], evaluate the effect of resistance training [47], and reveal muscle morphological decline of the supraspinatus muscle due to a shoulder injury [48]. Ultrasound was more spatially localized than EMG. The diameter of the ultrasound sensor used in this study was only 5 mm. The muscle thickness changes were calculated by detecting the shift of the reflected ultrasound echoes from the fat-muscle and muscle-bone interfaces. Therefore, the 1-D SMG signal provides the potential advantage of being able to noninvasively detect muscle thickness changes at different depths or locations with a single ultrasound transducer, which effectively avoids adjacent muscle cross talk. Compared with the MMG signal, the 1-D SMG signal was more stable and less influenced by external noise, such as movement artifact or power supply noise. In addition, the magnitudes of both EMG and MMG signals can be significantly affected by changes in the coupling conditions between the skin surface and the electrodes/sensors. Body motion can easily induce such changes in coupling condition, resulting in the distortion of EMG or MMG signals. In the case of 1-D SMG, the coupling between the ultrasound transducer and the skin can also be potentially affected by body motion. However, such effects may only affect the magnitude of the ultrasound echoes not the time between echoes, which represent muscle thickness. For this reason, 1-D SMG may be more robust to motion artifacts compared with EMG. Therefore, we expect that the 1-D SMG signal can be used as a complementary tool to EMG and MMG and provide muscle thickness information during various muscle contractions.

We demonstrated in this study that the 1-D SMG signal can be a complementary approach to surface EMG for detecting muscle activities and predicting the motions of individual skeletal muscles. Therefore, according to the current results, we believe that our 1-D SMG system has the potential to assess muscle function, muscle fatigue, and muscle pathology and control a prosthesis. However, the advantages of 1-D SMG should be further verified together with other factors that may affect the collection and extraction of the 1-D SMG signals.

Although the current experiment showed that 1-D SMG is a promising method for muscle assessment, it had some limitations. In this preliminary study, the 1-D SMG signal was found to be significantly correlated with wrist extension angle when no external resistance was present during the wrist extension movement. However, in future studies, the condition when external resistance is present should be investigated. In addition, skeletal muscles usually work as a group to perform certain movements. In this experiment, only the extensor carpi radialis muscle was studied during the wrist extension movement. Future investigators may hope to detect coactivated muscles by

using a multiple sensor SMG system. Finally, we found that placing both the 1-D SMG and surface EMG sensors on their optimal sites was difficult when the superficial area of the muscle was small. This problem should also be addressed in further feasibility tests of 1-D SMG.

In summary, this study demonstrated a significant linear relationship between the 1-D SMG and wrist angle signals. However, this is only a preliminary study of the use of 1-D SMG for detecting morphological changes of skeletal muscles. Whether such a correlation exists in other body joints or in disabled subjects requires further study. We also expect that 1-D SMG could be used along with surface EMG to provide more comprehensive information about skeletal muscle contractions. The correlation and differences between 1-D SMG and surface EMG should be further studied in future experiments.

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The authors have declared that no competing interests exist.

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