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Evaluation of the Power Assist Effect of Muscle Suit for Lower Back Support

MIYU IDE, TAKUYA HASHIMOTO, (Member, IEEE), KENTA MATSUMOTO, AND HIROSHI KOBAYASHI, (Member, IEEE)
Department of Mechanical Engineering, Tokyo University of Science, Tokyo 162-8601, Japan
Corresponding author: Takuya Hashimoto (tak@rs.tus.ac.jp)

ABSTRACT In order to reduce the burden on the lower back, we have been developing a compact and lightweight ''Muscle Suit'' that uses McKibben artificial muscles, a type of soft pneumatic actuator. The Muscle Suit for lower back assist has been commercialized and sold over 13,000 units so far; the evaluation of the assist effect is a key point for further development and marketing of the Muscle Suit. In this study, we focus on the assistive effects of two Muscle Suit models: the standard model and the standalone model. The former generates assistive force actively using the actuator, but the latter equips a mechanism generating assistive force passively without the actuator. In the experiments, we first conducted surface electromyogram (sEMG) measurement to examine muscle usage in lifting motion of a heavy weight with or without the Muscle Suit assists. Besides, we estimated the assist force of the Muscle Suit from the perspective of muscle usage ratio. As the result, the evaluation of the overall muscle usage of all measured muscles revealed that the both models of the Muscle Suit reduced muscle usage. We then examined whether the dynamic length of body sways (DLNG) could be used as an indicator of fatigue progressed with performing the lifting motion repeatedly under assist or non-assist conditions. The experimental results confirmed that the reduction of physical fatigue by the Muscle Suit suppressed the change of DLNG value.

INDEX TERMS Assistive technology, fatigue, length of body sways, lower back support, muscle suit, muscle usage.

I. INTRODUCTION

With the advances in science and technology, industrial robots and transportation equipment have replaced much of the heavy muscular work performed by humans; hence, the number of instances of work involving extreme physical burden is decreasing. However, there are cases of burdensome heavy work that are still difficult to mechanize or automate from the standpoint of work content and environment. Some examples include diaper changing, body posture changing, and transfer assistance for long-term care site residents, as well as transportation and sorting in construction and logistics sites with limited space. Many work-related injuries such as lower back pain still occur in places like long-term care sites, even though measures to prevent them are extensively promoted in Japan. Lower back pain currently accounts for a large percentage of such work-related disorders in many industries. According to a report published by the Ministry of Health, Labour, and Welfare [1] in 2017, of the 7,844 cases of work-related injuries that needed at least four days of leave of absence, 5,051 cases concerned lower back pain due to injury or work-related activities, accounting for approximately 65% of the total. Moreover, during the 10-year period starting from 2003, such cases have increased 2.7-fold in the public health and hygiene industry, which includes social welfare facilities. Therefore, it can be agreed that implementing health-promotion measures to prevent lower back pain is still an important task. In policy regarding lower back pain prevention in workplaces, one measure suggested by the Ministry of Health, Labour, and Welfare [2] is to reduce work that imposes a large burden on the lower back, such as lifting heavy objects, holding and transferring patients and elders, and working in an unnatural posture, among other examples. It must be noted that in the USA and Europe, guidelines for measures against lower back pain have been formulated [3], [4].

With respect to such lower back pain, a variety of wearable devices for lumbar support have been developed so far [5]–[10], and some types of back-support exoskeleton are commercially available for mainly industrial purposes

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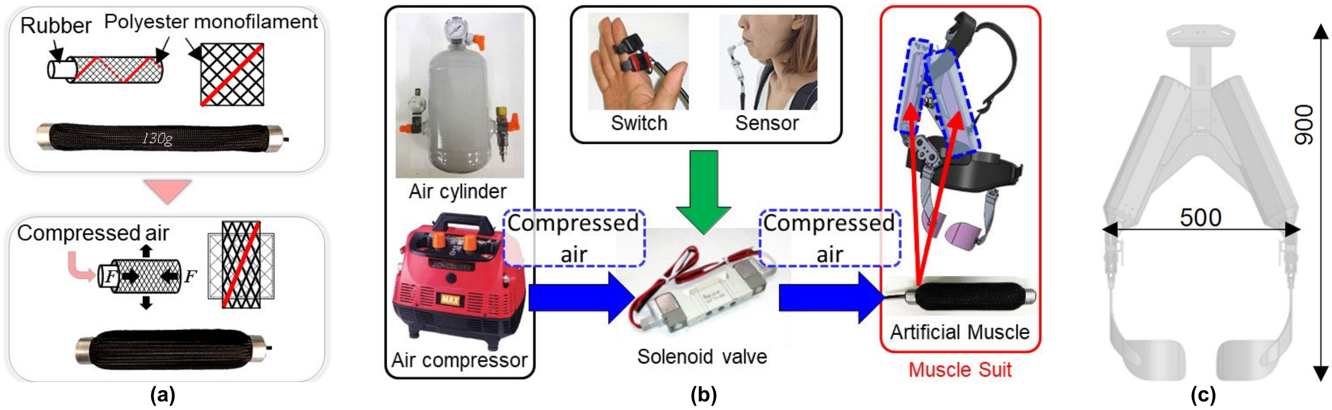

FIGURE 1. Main components for lower back support with the Muscle Suit. (a) McKibben-type artificial muscle. (b) Hardware equipment for controlling the Muscle Suit. (c) CAD model of the Muscle Suit and its size.

[11]–[13]. The authors also have been developing Muscle Suit®(Fig. 1), a wearable muscle support device that physically assists human motion [14]–[17]. Moreover, Innophys, a start-up affiliated with Tokyo University of Science, has been marketing Muscle Suit for lower back assist [18] since September, 2014, selling more than 13,000 units so far (as of July, 2020).

On the other hand, to increase the adoption of such wearable robots, it is important not only to promote the actual use in the work environment, but also to quantitatively evaluate the effectiveness of wearing such devices. Accordingly, several studies used surface electromyographic (sEMG) data to evaluate the effect of exoskeleton use [19]–[21]. Besides, with the Muscle Suit, the authors have previously evaluated the muscle usage through electromyogram measurements [22], as well as the muscle fatigue through near-infrared spectroscopy [23], [24]. However, these studies have mainly focused on muscle usage and muscle fatigue locally in the lower back. Even though the use of exoskeleton involves forced body motion affecting muscle coordination [25] and body center of gravity shift, the effect on the whole body has not yet been adequately taken into consideration. In this study, we first evaluated the effect of the Muscle Suit using electromyography as an ordinary manner and estimated the assist force from the perspective of muscle usage. However, the effect of the Muscle Suit was not obtained at some target muscles because the ratio at which the muscles are used differed for different conditions. Hence, we introduced a metric to evaluate the overall muscle usage taking into consideration the differences in muscle usage ratios. We then investigated a method to quantitatively evaluate the assist effect of the Muscle Suit based on the degree of fatigue in the entire body as a proposed method. The fatigue was measured using the total length of the body sway (LNG) at the center of gravity [26], [27]. In particular, this study proposed a novel evaluation metric, the dynamic length of body sways (DLNG), which is the LNG during motion. In both experiments, two types of the Muscle Suit were compared: standard and standalone models. In the standard model, the assist force is actively generated by an external supply of compressed air; in contrast, in the standalone model, the assist force is passively generated.

II. OVERVIEW OF THE MUSCLE SUIT FOR LOWER BACK SUPPORT

A. STANDARD MODEL

1) ACTUATOR

The Muscle Suit uses McKibben artificial muscles [28], [29], as actuators. Fig. 1(a) shows the structure and operating mechanism of this type of artificial muscle. The McKibben artificial muscle consists of a rubber tube caulked at both ends. This tube is covered with a sleeve based on a woven lattice made of polyester monofilament fiber, which has low elasticity. When compressed air is injected into the rubber tube, the pressure causes the artificial muscle to expand radially, decreasing the overall length. At this point, the changes in the fiber pattern angle in the sleeve generate a strong longitudinal constrictive force. With a simple and lightweight structure, it has a large output per unit weight characteristic compared to other actuators such as an electric motor. Moreover, it is possible to smoothly assist the human body motion since compressible air is used. The McKibben type artificial muscles used in the Muscle Suit for lower back assist have a diameter of 1.5 inches, a natural length of 300 mm, and weight of 130 g, and can generate a maximum tensile force of approximately 2,200 N using a (compressed air) supply of 0.5 MPa.

2) SYSTEM CONFIGURATION

The system configuration of the standard model of the Muscle Suit for lower back assist is shown in Fig. 1(b). It consists of the Muscle Suit itself, a compressed air source such as an air compressor, a solenoid valve for supplying and exhausting air for the artificial muscle, and a sensor or switch to control the solenoid valve. In the commercial version of the product, two switch types are available: a breath-activated switch, which the wearer can control by inhaling or exhaling into the

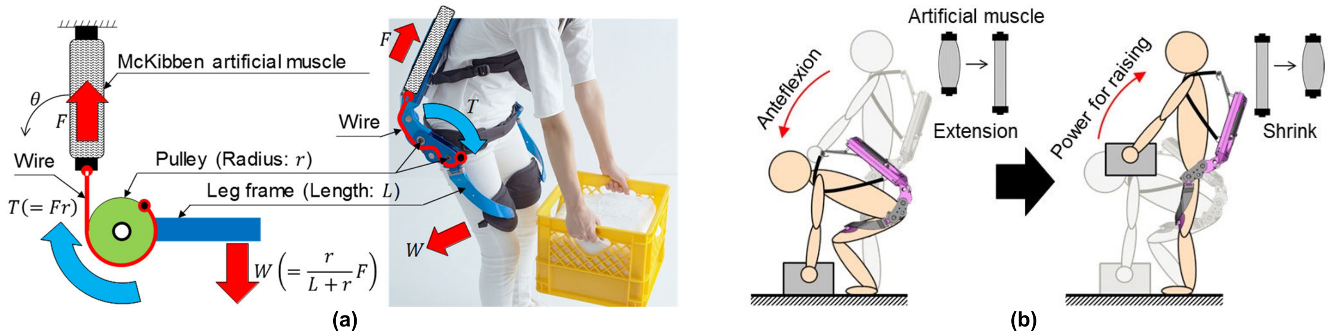

FIGURE 2. Conceptual mechanism for lower back support with the Muscle Suit. (a) Standard model generating active assistive force. (b) Standalone model generating passive assistive force.

mouthpiece, and a touch sensor switch placed on the chest that can control the valve when touched with the chin.

3) OPERATING PRINCIPLE

The structure of the standard model of the Muscle Suit for lower back assist is shown in Fig. 1(c). It has a height of 900 mm, a width of 500 mm, a depth of 220 mm, and a weight of 5.5 kg. The artificial muscle is fixed to the back frame on its upper end, and a wire is attached to the lower end connected to a pulley. A constrictive force is generated in the longitudinal direction by injecting compressed air into the artificial muscle. This force is then transmitted to the pulley through the attached wire, thus transforming it into the rotational force acting on the back frame in order to raise the upper body. Moreover, the leg frame extends from the pulley to the thigh pad, which covers the front part of the thigh, and receives the reaction force when raising the upper body. In this mechanism shown in Fig. 2(a), the torque rotates the upper body with respect to the thigh. In this way, the assist force facilitates the straightening of the thigh and the upper body, thus extending the lower back. Accordingly, assist forces are generated in both cases, when raising the upper body with the thigh in the upright posture, or when lowering the waist while keeping the upper body in an upright posture for lifting an object using the power of the leg. Consequently, the burden on the lower back and the legs is reduced.

The maximum weight handled by the Muscle Suit for lower back assist was set as 30 kgf, according to Chapter 6 of the Labour Standards Act (Sanitation Standards). To achieve an assist force of 30 kgf, an assist torque of 120 Nm must be generated in order to raise the upper body with respect to the lower body. Even if the 30 kgf (approximately 300 N) is intensively loaded on the upper body, this magnitude was used as a target measure for the assist force. In addition, to achieve this torque, a total of four McKibben type artificial muscles were used, two of them on each side. In the actual measurements of the extension torque, a maximum of 140 Nm was generated using a supply of 0.5 MPa (details described below).

B. STANDALONE MODEL

In the standard model, the artificial muscle is actively operated by an external supply of compressed air. Accordingly, an air compressor or an air tank, and a tube to supply compressed air are required. Moreover, as described previously, a breath-activated switch and a touch sensor switch are installed to enable the wearer to switch on and off the assist power. One reason for the manual operation by the wearer, is that it is difficult to automatically detect his or her intent of motion. Another reason is that, rather than performing the assist after the wearer starts to move, the assistive force is easily transmitted when the wearer adjusts the body to the movements of the Muscle Suit once it has started to operate. In addition, it has also been found that it is difficult for the user to operate the switches, and that the compressor itself and the tube extending from the compressor are hindrances.

Accordingly, a standalone model where switches and compressors are not required was developed. The underlying principle, that is, the method to move the artificial muscle without supplying compressed air externally, is shown in Fig. 2(b). First, air under a certain pressure P_{ini} [MPa] is supplied to the artificial muscle when the wearer is standing upright, and then the air nozzles are shut off. Under such a condition, when the wearer leans forward, as shown in Fig. 2(b), the artificial muscle extends in the longitudinal direction, thus reducing the volume to approximately 40% of the initial volume at P_{ini}. Since the air nozzles of the artificial muscle are shut off, according to the Boyle-Charles law (the product of pressure and volume is constant), the pressure inside the artificial muscle is raised to approximately $2.5\times P_{ini}$ [MPa]. Therefore, an assist force is generated like the case of supplying compressed air externally at $2.5\times P_{ini}$ [MPa]. It should be noted that a certain amount of force is necessary here for the wearer to lean forward. However, since the force can be generated by applying the body weight, it does not cause a big burden on the user. Using this principle, the assist force can be obtained simply by changing the motion direction of the wearer, thus making the switches unnecessary.

Since the standalone model has been marketed, nearly 100% of existing users have preferred it over the standard model.

C. OUTPUT TORQUE

Fig. 3 shows the output torque T [Nm] for the standard model (solid line), and the standalone model (dotted line).

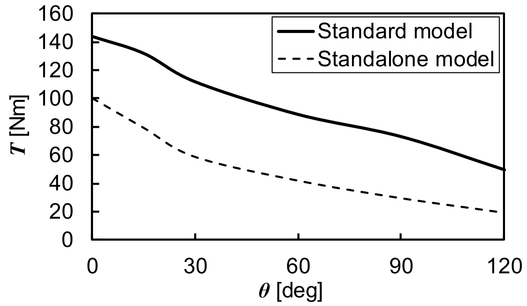

FIGURE 3. **Output torque characteristics of the standard model and the standalone model.**

The horizontal axis shows the angle θ [deg] of the Muscle Suit, as depicted in Fig. 2. If the back frame is at a horizontal position, corresponding to the Muscle Suit being bent forward, the measured angle is taken as $\theta = 0$ deg; hence, the state at which the wearer is standing upright is given by $\theta = 120$ deg. For the standard model, the torque corresponds to the case where the compressed air at 0.5 MPa was supplied to the artificial muscle. For the standalone model, the torque corresponds to the case where the compressed air at $P_{ini} =$ 0.12 MPa was supplied to the artificial muscle when the wearer was standing upright ($\theta = 120$ deg). This 0.12 MPa value was set as the initial pressure, based on the following observation. It was found that, while it varies among subjects, values higher than this increase the difficulty for the wearer to lean forward to the horizontal position. The results in the plot show that a maximum output torque of approximately 140 Nm for the standard model, and a maximum output torque of approximately 100 Nm for the standalone model were achieved. Please note that, since the value of P_{ini} corresponds to the maximum output torque, the user can adjust this value as necessary using devices such as a hand-operated pump.

III. EXPERIMENT 1: EALUATION OF MUSCLE USAGE USING SURFACE ELECTROMYOGRAM

To verify the assist effect of the standard and the standalone models of the Muscle Suit for lower back assist, the differences in muscle usage in the presence and absence of the device were studied using experiments involving lifting of weights. In the literature, ElectroMyoGraphy (EMG) is commonly used as a method for estimating physical burden such as local usage and muscle fatigue. Among the related methods, the surface ElectroMyoGram (sEMG), which is a noninvasive method, is the most commonly used [30], [31]. Accordingly, this study used sEMG method to quantitatively evaluate the muscle usage. In the experiments, the assist effect was determined to be present, if the muscle usage was reduced by using the device when lifting the same load (weight). Moreover, in the presence of the device, the load that showed the same muscle usage as in the absence of the device was determined to estimate the assist force from the muscle usage viewpoint.

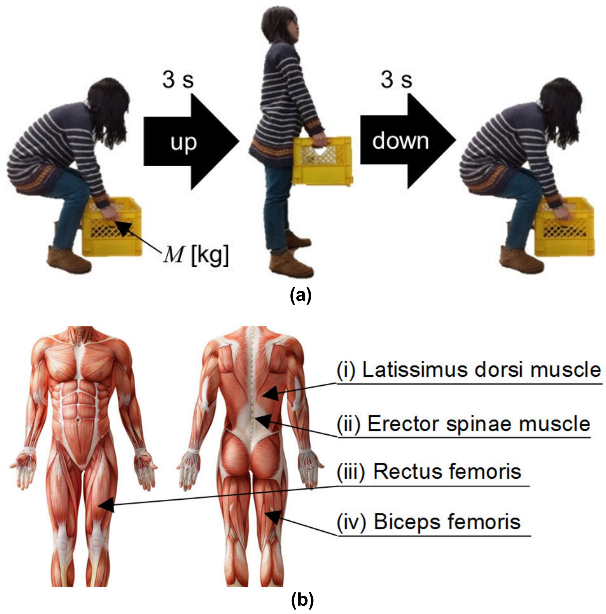

FIGURE 4. **Experimental condition. (a) Subject picks a weight up from the floor in the first 3 s and places it down on the floor in the following 3s. (b) Measurement points corresponding to four kinds of muscles involved in the lifting motion.**

TABLE 1. **Physical parameters of subjects in experiment 1.**

	A	B	C	D	Avg.
Gender	M	F	M	M	-
Height (cm)	171	150	170	165	164.0±9.7
Weight (kg)	60	45	60	70	58.8±10.3
Age (yrs)	22	23	23	23	22.8±0.5

A. EXPERIMENTAL METHOD

As shown in Fig. 4(a), the motion defined for the experiment consisted in picking up a load of M [Kg] from the floor in the first 3 s and placing it back down on the floor in the following 3 s. For the standard model, once the subject's hand grabbed the weight on the floor, compressed air at 0.5 MPa was supplied to assist the upright motion of the body to lift the weight; afterward, the air was exhausted to assist in placing the weight down on the floor. In this way, by letting the body adjust to the Muscle Suit, the weight could be lowered without any difficulty. In the standalone model, compressed air at 0.12 MPa was supplied when standing upright, and then sealed. As mentioned previously, the assist forces were generated in the direction of motion to make the body upright, by crouching down to reach the weight placed on the floor. In the experiment, the defined motion was performed under three different assist conditions: (A) without any Muscle Suit assist, (B) with the standard model Muscle Suit (denoted as MS in the figure) assist, and (C) with the standalone model Muscle Suit (denoted as SA in the figure) assist. Besides, each trial was conducted under different weight (load) conditions: $M = 10$, 15, 20, and 25 kg.

As shown in Fig. 4(b), the electromyograms were measured at a total of eight points on the left and right sides of the

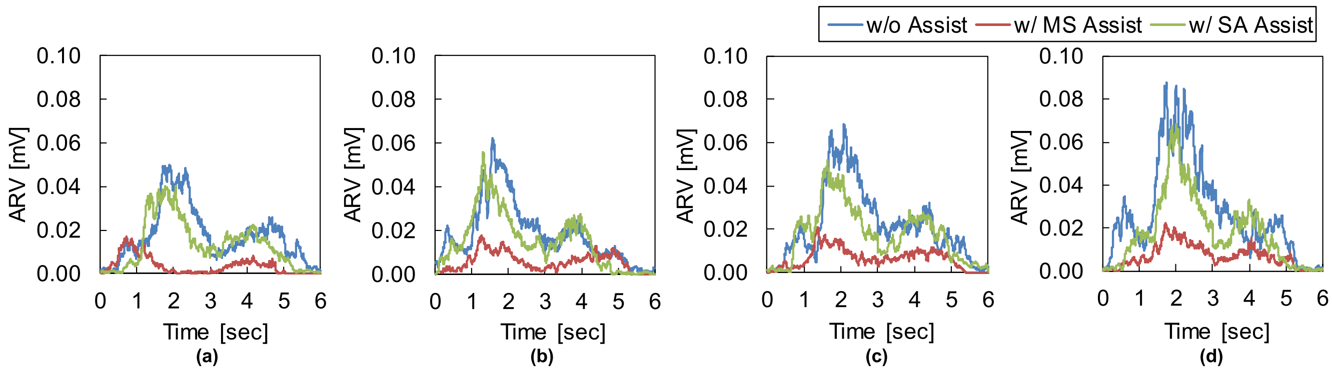

FIGURE 5. Examples of ARV waveforms derived from the measured sEMG signals of Subject D. (a) M = 10 kg. (b) M = 15 kg. (c) M = 20 kg. (d) M = 25 kg.

body on the following four muscles: (i) latissimus dorsi muscle, (ii) erector spinae muscle, (iii) rectus femoris, and (iv) biceps femoris. These muscles are thought to be used during the motion of lifting a weight. While lifting the weight, it is likely that for each of the experiments, the balance between the right and left sides of the body differed. Accordingly, the average evaluation values derived from sEMG signals on the left- and the right-side points were used for each of the muscles. The sEMG measurement was undertaken using a WEB-7000 (Nihon Kohden Co.), with a sampling frequency of 1 kHz, and a time constant of 0.1 s. Moreover, a 30 Hz low-pass filter, and a 500 Hz high-pass filter were applied.

The subjects were four healthy adults (A, B, C, D) in their twenties. Before conducting the experiments, the experimental content and associated risks were adequately explained to the subjects, and consent on participation in the experiments was obtained. The physical parameters of the subjects are summarized in Table 1.

B. DATA PROCESSING

Since the muscle activity becomes vigorous when the muscle usage increases, the corresponding average rectified value (ARV) and integral electromyogram (IEMG) of the EMG also increase. Accordingly, ARV and IEMG were used as the indicators of muscle usage in this study. ARV and IEMG were calculated using the following equations (1) and (2), respectively,

$$ARV\,(t) = \frac{1}{2T}\int_{-T}^{T} |S\,(t+\tau)|\,d\tau[\text{mV}] \tag{1}$$

$$IEMG = S = \sum ARV(t)\Delta t[\text{mV}\cdot\text{s}] \tag{2}$$

where τ is a time constant, and $(-T, T)$ is the calculation interval. For the sEMG waveform measurements at each of the points, the following post-processing was undertaken:

a) Generating ARV waveform by rectifying and smoothing the EMG waveform
b) Averaging ARV waveform for each of the muscle points measured on the right and left sides
c) Calculating IEMG for each of the points measured under each assist condition and each load condition

TABLE 2. Estimation results of assist force of the two models.

Sub.	(B) w/ MS Assist		(C) w/ SA Assist	
	M'_{MS}	$M'_{MS}-M_{w/o}$	M'_{SA}	$M'_{MS}-M_{w/o}$
A	20	10	15	5
B	20	10	20	10
C	25	15	20	10
D	25	15	25	15

Unit: kg

C. EXPERIMENTAL RESULTS

As one example of the experimental results, the ARV waveforms of the latissimus dorsi muscle under each of the conditions for subject D are shown in Fig. 5. Moreover, Fig. 6 shows the result of IEMG calculations for all the subjects under all the conditions. In the description of the figures hereafter, the following abbreviations are used: (A) w/o Assist for the condition without any Muscle Suit assist, (B) w/ MS Assist for the condition with the standard model Muscle Suit assist, and (C) w/ SA Assist for the condition with the standalone model Muscle Suit assist. These results are discussed and verified as follows.

1) RELATIONSHIP BETWEEN LOAD AND MUSCLE USAGE

As can be observed in Figs. 5 and 6, for most of the experimental conditions, the ARV and IEMG values increased with the increase in load. It is likely that this was caused by the burden on the body increasing with the increase in load, requiring the use of more muscles. However, as shown in Fig. 6(c) for subject C, there were cases where the IEMG did not increase monotonously for the (ii) erector spinae muscle under the assist condition with SA model, as shown in frame 1 marked in red. This is likely caused by the usage ratios of the different muscles differing in the lifting motions for different weights.

2) COMPARISON OF MUSCLE USAGE FOR THE SAME LOAD WITH OR WITHOUT ASSIST

How the muscle usage varies with or without the assist force for the same load is verified in this section. According to

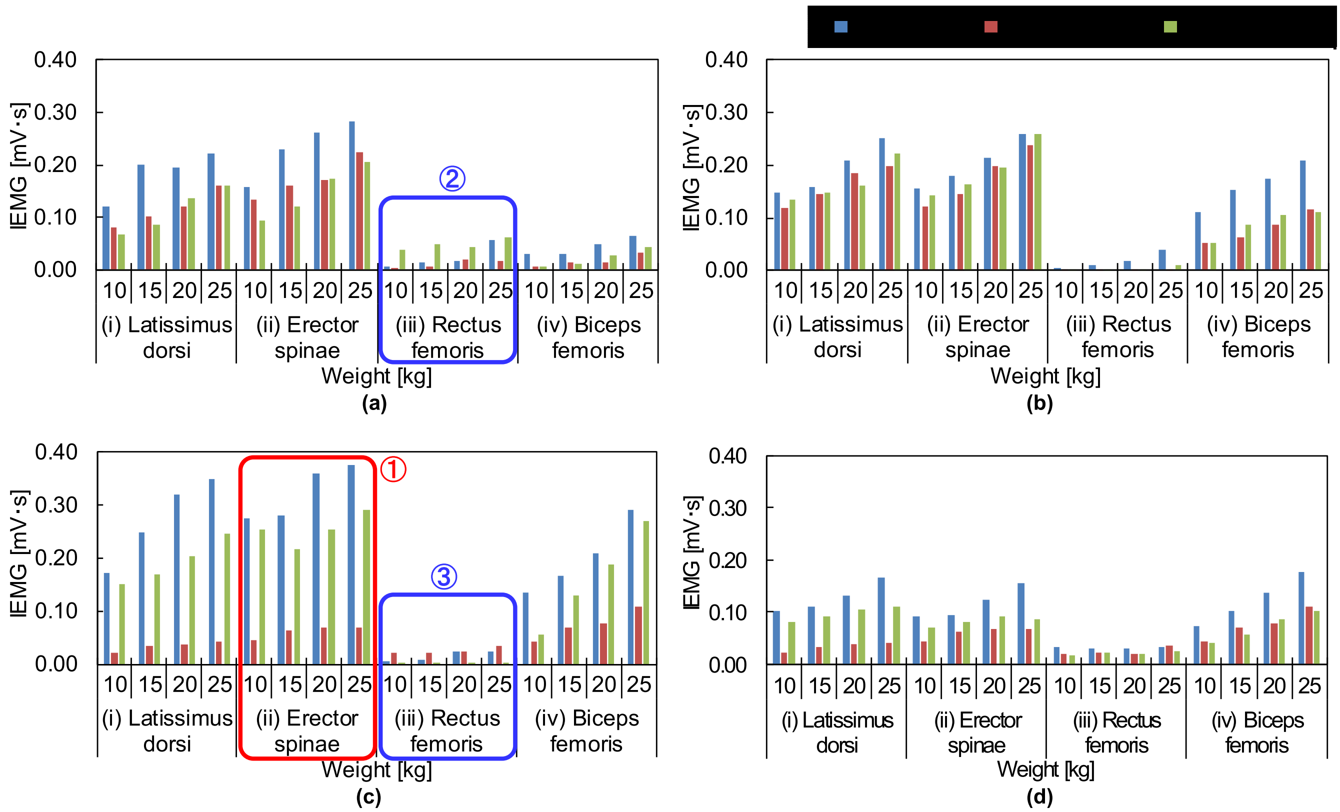

FIGURE 6. **Results of IEMG calculation in each subjective muscle. (a) Subject A. (b) Subject B. (c) Subject C. (d) Subject D.**

Figs. 5 and 6, for many of the load conditions under both conditions (B) w/ MS Assist and (C) w/ SA Assist, the ARV and IEMG values become smaller compared to those for condition (A) w/o Assist. However, for subject A under condition (C) w/ SA Assist, as shown in Fig. 6(a) in frame 2 marked in blue, and for subject C under condition (B) w/ MS Assist, as shown in Fig. 6(c) in frame 3 marked in blue, the IEMG values for the rectus femoris were larger than those under the condition (A) w/o Assist.

3) EVALUATION OF MUSCLE USAGE USING MUSCLE USAGE RATIO

The discussions in 1) and 2) indicate that it is likely that depending on the condition, the muscle usage ratios differ. Keeping this in mind, the overall muscle usage of all measurement points was evaluated taking into consideration the differences in muscle usage ratios. The following procedure was performed for evaluating the overall muscle usage. First, the IEMG values were denoted by $S(i, j, k)$, for the weights (load) as $M_i = 10, 15, 20, 25$ kg ($i = 1, 2, 3, 4$), the assist conditions as $a_j =$ (A), (B), (C) ($j = 1, 2, 3$), and the measurement points as $p_k =$ (i), (ii), (iii), (iv) ($k = 1, 2, 3, 4$). For example, $S(1, 1, 1)$ shows the IEMG value for load $M_1 = 10$ kg, lifting motion assist condition $a_1 =$ (A): without Muscle Suit assist, and point $p_1 =$ (i): latissimus dorsi muscle. For the four measurement points, the IEMG average values ($= \bar{S}(k)$), as given by equation (3), were calculated for all the load conditions and all the assist conditions. Next, using equation (4), the IEMG values for each point under each load and under each assist condition were divided by the average value calculated in the previous step to obtain the standardized IEMG value ($= S'(i,j,k)$ for each point. Lastly, using equation (5), the sum of the standardized IEMG values ($= S''(i,j)$) (hereafter, SS-IEMG) for the four points was calculated and used as the representative value for each of the assist conditions (A)–(C) and for each load.

$$\bar{S}(k) = \frac{1}{12}\sum_{i=1}^{4}\sum_{j=1}^{3} S(i, j, k) \tag{3}$$

$$S'(i, j, k) = \frac{S(i, j, k)}{\bar{S}(k)} \tag{4}$$

$$S''(i, j) = \sum_{k=1}^{4} S'(i, j, k) \tag{5}$$

The results obtained using this procedure are shown in Fig. 7. The results shown in Fig. 7(a)–(d) indicate that for all the subjects, under all the assist conditions, the SS-IEMG value increased proportionally to the load. In this way, the muscle usage can be comprehensively evaluated by measuring the muscle usage at all the points, even if the whole-body motion and the muscle usage ratios differ at times. It can also be observed that, as the physical burden on

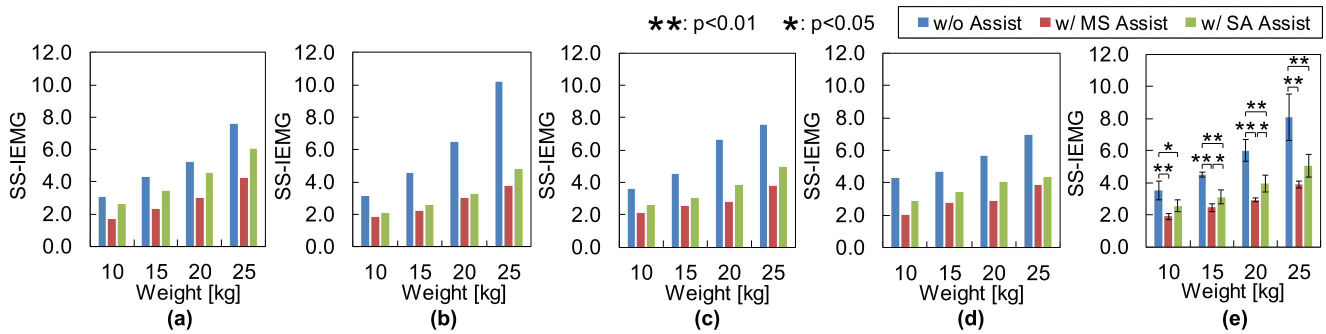

FIGURE 7. Results of SS-IEMG in each load condition. (a) Subject A. (b) Subject B. (c) Subject C. (d) Subject D. (e) Mean.

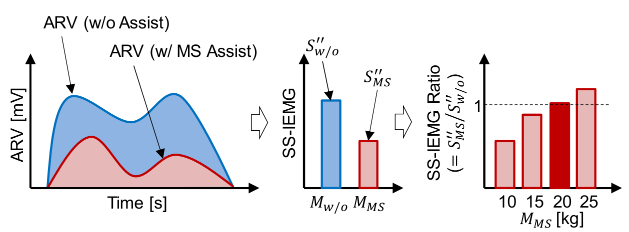

FIGURE 8. Comparison of SS-IEMG.

the body increases with the increase in load, the overall muscle usage also increases. Moreover, comparing the SS-IEMG values under assist conditions (A)–(C) for the same load, the values under w/ MS Assist and w/ SA Assist conditions are smaller than those under the w/o Assist condition for all the subjects. Fig. 7(e) shows the mean values of all subjects, and Tukey's test detects significant difference between the non-assist condition and the two assist conditions. Accordingly, the results indicate that for both the standard and standalone models, reduction in muscle usage was realized through the assist effect, lessening the burden on the body.

4) CALCULATION OF THE ASSIST FORCE THROUGH COMPARISON OF MUSCLE USAGE UNDER DIFFERENT LOADS

The discussions in 3) suggest that larger muscle usage correspond to an increased burden on the body. Accordingly, regardless of the presence or absence of the assist device, it can be considered that if the SS-IEMG values are similar, the corresponding burdens on the body are also similar. Keeping this in mind, for the conditions (B) w/ MS Assist, and (C) w/ SA Assist, the SS-IEMG values corresponding to the lifting motion for the different weights (loads) were compared. Next, the assist force for each model was verified by investigating the differences in the weights needed to induce a similar burden (muscle usage) on the body. Here, for the (A) w/o Assist condition, the weight was set as $M_{w/o} = 10$ kg, and for the (B) w/ MS Assist and (C) w/ SA Assist conditions, the weights were set as $M_{MS} = M_{SA} = 10$, 15, 20, 25 kg. The weights M_{MS}, M_{SA} [kg] for the assist conditions producing SS-IEMG values closest to that for $M_{w/o}$ [kg] were considered to be producing similar burden on the body and were set as M'_{MS}, M'_{SA} [kg]. The Muscle Suit assist force for the standard model was then calculated as M'_{MS}–$M_{w/o}$ [kg] and that for the standalone model was calculated as M'_{SA}–$M_{w/o}$ [kg].

The comparison method for SS-IEMG values is shown in Fig. 8. The SS-IEMG value corresponding to $M_{w/o}$ [kg] was taken as $S''_{w/o}\left(M_{w/o}\right)$, and the SS-IEMG values corresponding to M_{MS}, M_{SA} under assist conditions were taken as $S''_{MS}\left(M_{MS}\right)$, $S''_{SA}\left(M_{SA}\right)$. Then, $S''_{MS}\left(M_{MS}\right)/S''_{w/o}\left(M_{w/o}\right)$ and $S''_{SA}\left(M_{SA}\right)/S''_{w/o}\left(M_{w/o}\right)$ was calculated, and the weights M'_{MS}, M'_{SA} [kg] for which these values were closest to 1, that is, when the values for with and without assist were similar, were estimated.

Fig. 9 shows the results of $S''_{MS}\left(M_{MS}\right)/S''_{w/o}\left(M_{w/o}\right)$ and $S''_{SA}\left(M_{SA}\right)/S''_{w/o}\left(M_{w/o}\right)$ for different weights using the method described above. Moreover, Table 2 shows the results of M'_{MS}–M'_{SA} [kg] such that the SS-IEMG ratio was closest to 1; these values were used to calculate M'_{MS}–$M_{w/o}$ [kg] and M'_{SA}–$M_{w/o}$ [kg].

Using Table 2, the average and standard deviation of the assist force were calculated. For the standard and the standalone models, these were 12.5 ± 2.5 kgf and 10.0 ± 3.5 kgf, respectively. Since the motion also involved the lifting of the upper bodyweight, the external assist force was affected by the weight of the upper body, and depending on the (bending) angle, and the effect of gravity, the resulting assist forces differed. On the other hand, as described in section II-C, the maximum output torque for the standard and standalone models was approximately 140 Nm and approximately 100 Nm, respectively. This proportion is approximately similar to that of 12.5 ± 2.5 kgf and 10.0 ± 3.5 kgf, suggesting that the results obtained are valid.

5) RELATIONSHIP BETWEEN RECTUS FEMORIS MUSCLE USAGE AND ASSIST FORCE

As described above, comparison of the assist forces for (B) the standard model and (C) the standalone model calculated from the verification of IEMG values at all the points, reveals large variations among the subjects. As shown in Fig. 6(a), for subject A where the assist force is the smallest, the IEMG values at the (iii) rectus femoris under the (C) w/ SA assist condition were rather larger than those under

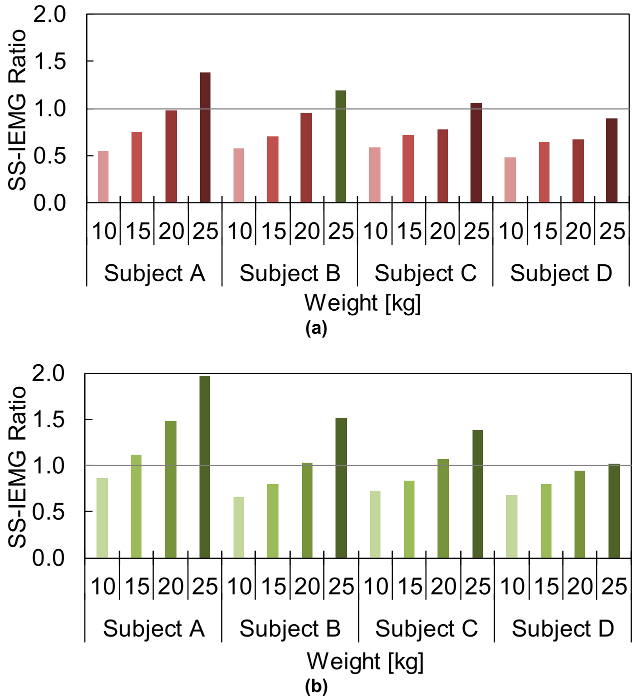

FIGURE 9. Ratios of SS-IEMG in the two assist conditions to that of the non-assist condition. (a) w/ MA Assist condition. (b) w/ SA Assist condition.

the (A) w/o Assist condition. Accordingly, it is likely that the IEMG values at all points increased, thus resulting in a lower assist force.

Rectus femoris is commonly used in the hip flexion, which is the operation when a person leans forward and bends the knees to crouch down. As explained in section II-B, in the standalone model, the assist force is generated through hip flexion when the wearer leans forward or crouches down by bending the knees. However, the wearer needs to apply force to crouch down. If the wearer can deftly use one's own body and is familiar with the standalone model, it is possible for the wearer to utilize the body weight and crouch down without using the power of the legs. However, if the wearer is not familiar with the device, then the power of the leg is also used while crouching down. Accordingly, it is likely that the muscle usage at the rectus femoris is increased, thus resulting in higher IEMG values at all the points, and lower values of the assist force.

In this way, in the case of the standalone model, individual differences may occur due to experience in the operation of the device and body use. Accordingly, depending on the subject, adequate assist force may not be generated. Because of this, we are currently developing a new device that facilitates the operation of crouching down, by reducing the assist force when the crouching down angle is large.

IV. EXPERIMENT 2: EVALUATION OF FATIGUE USING LENGTH OF BODY SWAY AT THE CENTER OF GRAVITY

The length of the trajectory along with the body center of gravity shifting within a certain time is called the length of body sway (LNG). In general, the LNG value when standing upright at rest is used as a reference, and it is known that this value increases with the increase in fatigue [26], [27]. In this study, the LNG was measured when in motion, instead of at rest, to verify whether the effect on fatigue during the use of the Muscle Suit assist could be evaluated. Note that, in the description that follows, the LNG while standing upright at rest is referred to as static LNG (SLNG), and the LNG while in motion is referred to as dynamic LNG (DLNG).

TABLE 3. Physical parameters of subjects in experiment 2.

	E	F	G	H	I	J	Avg.
Gender	M	M	M	F	M	M	-
Height (cm)	167	177	171	150	166	170	164.0±8.4
Weight (kg)	65	59	60	45	63	55	58.8±8.9
Age (yrs)	23	24	22	23	24	25	22.8±0.4

A. MEASUREMENT METHOD OF DLNG

A force plate TF-4060-D-Fz (Tec Gihan Co., Ltd.) was used for the center of gravity measurements. The sampling frequency of the measuring device was 100 Hz. However, it is known that the average human reaction time is approximately 0.19 s (5 Hz) [32], and assuming a maximum reaction time of 0.1 s (10 Hz), the frequency of the center of gravity coordinates of a human being can be recorded as 10 Hz. Moreover, according to the sampling theorem, it is necessary to sample at more than twice the frequency of a given analog waveform to accurately restore it from its converted digital waveform. Accordingly, in this study, the center of gravity coordinate data obtained at 100 Hz was smoothed at 20 Hz; two times 10 Hz.

1) INVESTIGATION OF MEASUREMENT DURATION FOR CALCULATING DLNG

LNG is the length of the body sway at the center of gravity, which needs to be measured over a certain time (Δt). In the case of DLNG, since the subject is always in motion, a suitable Δt needs to be investigated. Accordingly, for the motion shown in Fig. 4(a) (6 seconds at a time), the experiments were repeated 20 times, with $M = 15$ kg, and three males (E, F, G) and one female (H) as subjects. The physical parameters of the subjects are summarized in Table 3. Considering that the motion at one time lasted 6 s, with $\Delta t = 6, 12, 18, 24, 30, 36, 42, 48, 54, 60$, the change of LNG over time (DLNG) within the period $[t - \Delta t, t]$ was studied. Fig. 10 shows one such example. The results indicate that, for a Δt greater than 30, it is easier to visualize the change trend. Since the fatigue increases as the motion is repeated, the smaller the Δt, the better the trend can be observed. Accordingly, $\Delta t = 30$ was used in this study. In the case of the SLNG, a value for standing upright at rest over 60 s is generally used, but 30 s was also used for the SLNG to be consistent with the DLNG.

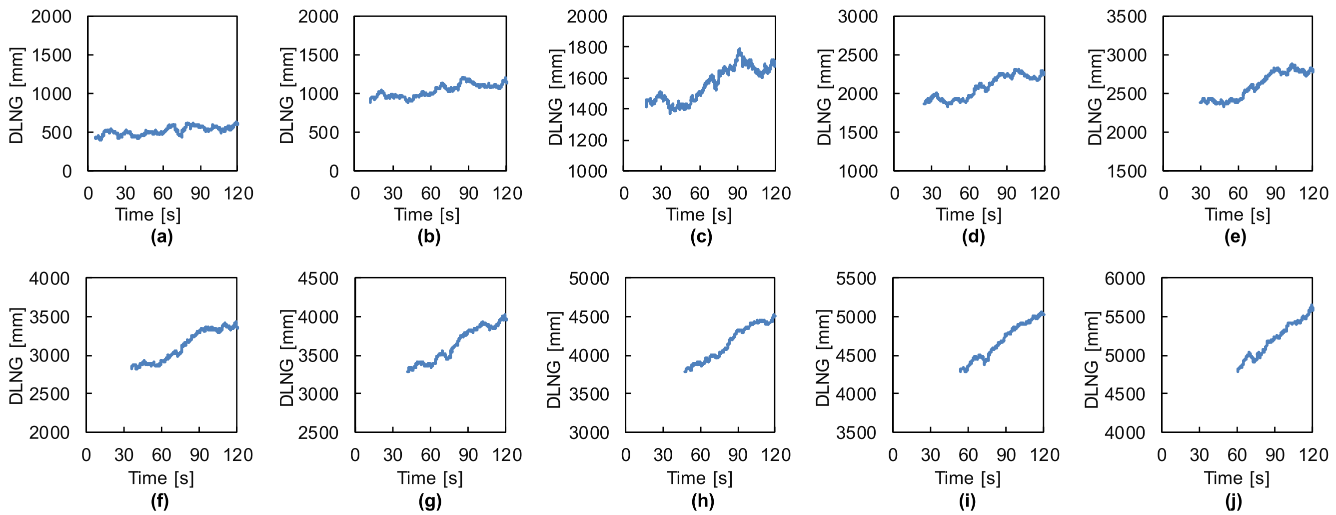

FIGURE 10. **Calculation results of dynamic LNG (DLNG) at different time interval. (a) Δt = 6 s. (b) Δt = 12 s. (c) Δt = 18 s. (d) Δt = 24 s. (e) Δt = 30 s. (f) Δt = 36 s. (g) Δt = 42 s. (h) Δt = 48 s. (i) Δt = 54 s. (j) Δt = 60 s.**

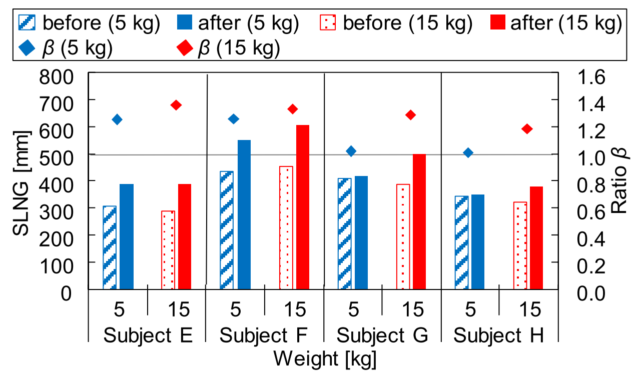

FIGURE 11. **Result of SLNG measurement.**

2) COMPARISON OF DLNG UNDER DIFFERENT FATIGUE CONDITION

The experimental motion described in section III-A was repeated 20 times under low ($M = 5$ kg) and high ($M = 15$ kg) load conditions to verify the DLNG trend due to fatigue, using SLNG as the known indicator of fatigue. The subjects were the same (E, F, G, H).

SLNG is generally used as an indicator of fatigue, but to confirm the change before and after the motion, the ratio of SLNG after the motion to that before the motion, set as β, was used here. A large β (>1.0) shows that the SLNG after the motion is greater than that before the motion, indicating an increase in fatigue. Fig. 11 shows the SLNG (mm) before and after the motion and the corresponding β for each subject. Moreover, the change DLNG over time is shown in Fig. 12.

The plots in Fig. 11 indicate that, for all the subjects' SLNG for standing upright at rest under both conditions, $\beta > 1$, and the β corresponding to the 15 kg case is greater than that of the 5 kg case. The overall impression about the experiments was that all the subjects experienced more fatigue for the 15 kg load than for the 5 kg. The fact that β for the 15 kg load was greater corroborates this impression. These results, using SLNG as an indicator, confirmed that for the lifting motion, a heavier weight (load) corresponds to increased levels of fatigue.

Moreover, the plots in Fig. 12 indicate that, under both load conditions, the DLNG value increases over time, and the increase ratio of DLNG is larger for the 15 kg case. These observations confirmed that, for both states of standing upright at rest and in motion, the LNG increases due to fatigue, and, as the fatigue accumulates (becomes larger), the LNG increase ratio also increases. Accordingly, the results suggest that a quantitative representation of fatigue using SLNG and DLNG is feasible.

B. EVALUATION OF THE ASSIST EFFECT OF THE MUSCLE SUIT USING DLNG

1) EXPERIMENTAL METHOD

It can be considered that the larger the assist effect of the Muscle Suit, the smaller the degree of fatigue due to repeated motion for lifting weights. Accordingly, DLNG was measured under the three conditions of (A) w/o Assist, (B) w/MS Assist, and (C) w/SA Assist, by repeating the experimental motion described in section III-A for 20 times with $M = 15$ kg. The degree of fatigue was then investigated by comparing the slopes of the DLNG. Four healthy adults (three males (G, I, J)), one female (H)) shown in Table 3 participated in this experiment.

2) EXPERIMENTAL RESULTS

The plots in Fig. 13 show the variation of DLNG over time. First, according to the interview survey on the experience, conducted after the experiments, all of the four subjects felt fatigued midway under the (A) w/o Assist condition. However, under the (B) w/ MS Assist and (C) w/ SA Assist conditions they did not feel much fatigue. Moreover, observing the

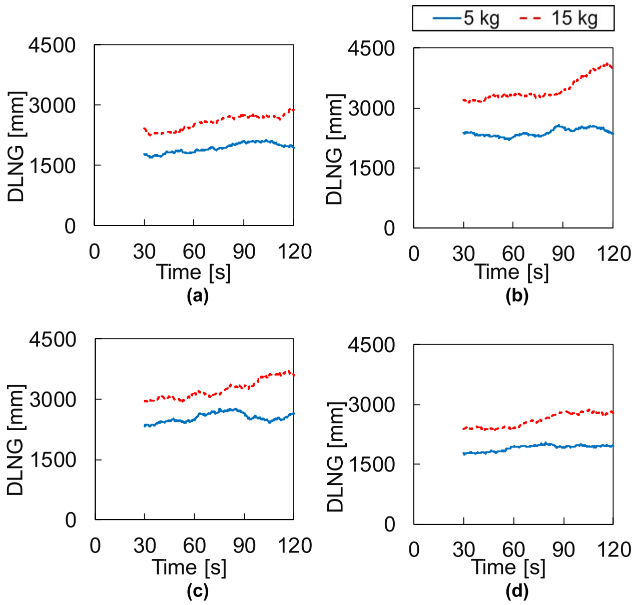

FIGURE 12. Result of DLNG measurement. (a) Subject E. (b) Subject F. (c) Subject G. (d) Subject H.

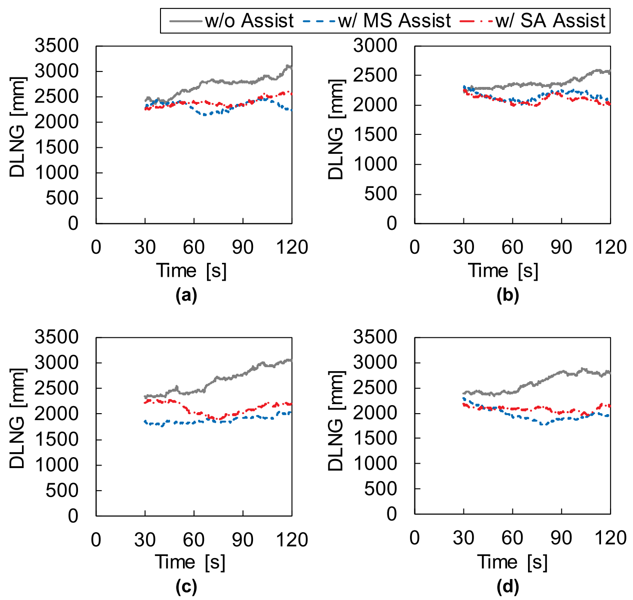

FIGURE 13. The results of DLNG measurements in different assist conditions. (a) Subject G. (b) Subject H. (c) Subject I. (d) Subject J.

entire motion period shown in Fig. 13, it can be confirmed that for both conditions, the fatigue was reduced due to assist. This occurred because the increase ratios of the values shown in the graphs were smaller compared to those under the no assist condition. On the other hand, in the first half of the motion period, it was observed that regardless of the assist effect, in some cases the increase ratio was negative. This was likely due to the instability of the center of gravity corresponding to the motion, since the body is not yet accustomed to the motion of lifting the weight immediately after starting the motion, and fatigue has not been generated yet.

Accordingly, it was assumed that the participant becomes accustomed to the lifting motion from the sixth time onwards. With this assumption, the DLNG slopes from the sixth time and beyond, under the three conditions of (A) w/o Assist, (B) w/ MS Assist, and (C) w/ SA Assist, were estimated using the least-squares method. These results are shown in Fig. 14(a). Moreover, since the slope changes constantly over time, to compare the increase ratios, under each of the (A)–(C) conditions, the sum of the slopes of the DLNG, as calculated from each of the instances of motion (6 s), was calculated (Fig. 14(b)).

While there are individual differences, the results shown in Fig. 14 suggest that, for all the subjects, the slopes of the DLNG, the sum of the slopes of the DLNG as calculated from each of the motion instances, and the increase ratios of the DLNG values under (B) w/ MS Assist and (C) w/ SA Assist conditions are smaller than those under the (A) w/o Assist condition. Besides, Fig. 14 also shows mean values of all subjects, and the significant difference between the non-assist condition and the two assist conditions is observed by Tukey's test. The results described above suggest that, the fatigue reduction due to the assist effect of the Muscle Suit standard and standalone models can be quantitatively expressed by using the DLNG increase ratio as an indicator, and by comparing the degree of fatigue associated with the lifting motion under different assist conditions.

V. DISCUSSION

In the first experiment, the assistive effect of two types of Muscle Suits—standard and standalone models—was confirmed using surface electromyography. The assistive effect in comparison with individual muscles was not always observed because the usage ratios of different muscles vary in different assist and load conditions. Hence, we proposed a metric: the sum of standardized IEMG (SS-IEMG) to measure overall muscle usage of all measurement muscles. The comparison of SS-IEMG showed a consistent effect of the Muscle Suit even in different conditions. As a result, the standard model of the Muscle Suit, which actively generates assistive force on the lumbar region, had the greatest reduction in muscle usage. Furthermore, the estimated passive assistive force of the standalone model was approximately 80% of that of the standard model. Hence, it is confirmed that the standalone model also provides a sufficient assistive effect. The proposed metric used for the evaluation of overall muscle usage of all measurement muscles can reduce the effect of individual muscle variability. Therefore, it is a useful method when the number of subjects is small, and there are multiple muscles to be measured. Additionally, increasing the number of subjects reduces the variability in the subjects and muscles. A possible future direction for this study is to increase the variations in number, gender, age, and body size of subjects, thus making the assistive effect of the Muscle Suit more generalized.

In the second experiment, the assistive effect of the Muscle Suit in reducing fatigue was evaluated. For this experiment,

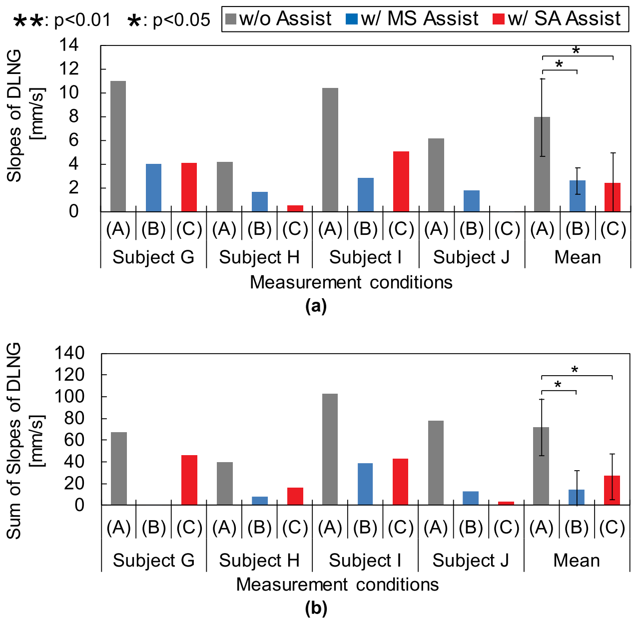

FIGURE 14. **Slope of DLNG waveform. (a) Slope of DLNG waveform from the sixth time and beyond using the least squares method. (b) Sum of the slopes of the DLNG calculated from each of the instances of motion (6 s).**

we proposed a novel metric: the dynamic length of body sway (DLNG), which represents the trajectory length of the center of gravity during movement. The result of DLNG measurement revealed that DLNG gradually increased during repetitive lifting motion without the Muscle Suit. Meanwhile, the increase in DLNG was suppressed with the Muscle Suit use. In literature, the body sway is related to fatigue [26], [27], [33], therefore, it is verified that the Muscle Suit decreases fatigue through its assistive effect. Furthermore, because the body sway is strongly associated with fatigue of the lower extremities, especially calf (ankle) muscles [34]–[36], the Muscle Suit has a potential to affect the strength of lower extremity muscles as well as that of dorsal muscles. In this study, only the thigh muscles were measured; hence, it is interesting to investigate the assistive effect of the Muscle Suit on the reduction of fatigue in more detail by measuring the other lower extremity muscles, including calf muscles. In addition, it would be important to investigate how the Muscle Suit affects the metabolic energy expenditure of the wearer from the perspective of overall muscle fatigue.

VI. CONCLUSION

In this study, the assist effect of the standard and standalone models of the Muscle Suit were quantitatively clarified. First, in section III, the reduction of burden on the body was confirmed through the assist effect of the standard and standalone models of the Muscle Suit for lower back support, estimated using surface electromyogram measurements. Moreover, comparing the IEMG values and estimating the weight (load) that produces similar burden on the body, the assist forces for the standard and standalone models were verified. In section IV, it was shown that, in addition to the static length of body sway (static LNG, or SLNG) as one of the conventional indicators of fatigue, the dynamic length of body sway while in motion (dynamic LNG, or DLNG) can also be used as an indicator of fatigue. Using these indicators, it was confirmed that the fatigue associated with performing repeated weight lifting motion can be reduced by the assist effect of the standard and standalone models.

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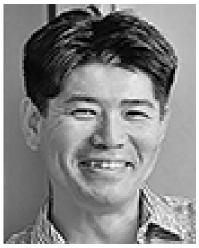
MIYU IDE received the B.S. and M.S. degrees in mechanical engineering from the Tokyo University of Science, Tokyo, Japan, in 2017 and 2019, respectively.

TAKUYA HASHIMOTO (Member, IEEE) received the B.S., M.S., and Ph.D. degrees in mechanical engineering from the Tokyo University of Science, Tokyo, Japan, in 2004, 2006, and 2009, respectively. He is currently a Junior Associate Professor with the Department of Mechanical Engineering, Tokyo University of Science. His research interests include robotics, human–robot interaction (HRI), HRI evaluation, sociability of robots, welfare robots, and human interface.

KENTA MATSUMOTO received the B.S., M.S., and Ph.D. degrees in mechanical engineering from Doshisha University, Kyoto, Japan, in 2013, 2015, and 2018, respectively. From 2018 to 2019, he was a Research Fellow with Doshisha University. He is currently an Associate Professor with the Department of Mechanical Engineering, Tokyo University of Science, Tokyo, Japan. His research interests include biomechanics, sports engineering, wearable device, and welfare robot.



HIROSHI KOBAYASHI (Member, IEEE) received the B.S., M.S., and Ph.D. degrees in mechanical engineering from the Tokyo University of Science, Tokyo, Japan, in 1990, 1992, and 1995, respectively.

From 1992 to 1995, he was a JSPS Postdoctoral Fellow with the Tokyo University of Science. From 1996 to 1998, he was a JSPS Postdoctoral Fellow for foreign study with Zurich University, Zurich, Switzerland. He was appointed as a Lecturer, in 1998, an Assistant Professor, in 1999, with the Department of Mechanical Engineering, Tokyo University of Science, where he has been a Professor, since 2008. He founded Innophys Company Ltd., in 2013, which is the manufacture and distributer of Muscle Suit®. His current research interests include practical robot systems, human interface, affective communication, welfare robot, intelligent image processing, and android robot.