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Paper:

Development and Control of Power-Assisted Lumbar Suit Based on Upper-Body Acceleration

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In this study, we propose a method to estimate the assistive timing requirements for a power-assisted lumbar suit based on upper-body acceleration. Our developed power-assisted suit combines of springs, wires, and an electrical motor to provide efficient assistance. The assistive torque provided by the suit was determined based on a digital human model. The assistive timing using the electrical motor was calculated from the upper-body acceleration measured using two internal accelerometers. Herein, we present the experimental results based on the myoelectricity of a muscle during lifting motions involving three participants acting as caregivers to elderly patients.

Keywords: power-assisted lumbar suit, nursing care, upper-body acceleration

1. Introduction

In recent years, the aging population in Japan has exceeded 28%, and the population of elderly care receivers correspondingly has increased [a]. In addition, the burden on caregivers has increased, which has become a significant problem. Many nursing care workers experience back pain because some nursing care motions place large strain on their lower backs [1]. Although floor lifts and sliding seats have been developed, facilities that can accommodate such equipment are limited [2–4].

Therefore, power-assisted suits are being actively researched and developed to reduce the burden on the lower backs of caregivers. The developed power-assisted suits can be broadly categorized into two types: active and passive. The active type provides assistance by using active elements such as electric motors, and can control arbitrary assistive power and timing parameters [5–7]. However, this type of suit is larger and heavier than the passive type, its operating time is limited by the battery life, and there is a trade-off between weight, drive time, and assistive output. In contrast, the passive type provides assistance using passive elements such as springs and rubber bands, and is not limited by the battery power level [8–10]. However, assistive power and timing are difficult to control with this type of suit. Therefore, we proposed and developed a power-assisted lumbar suit that combines active and passive elements [11, 12]. By incorporating a spring mechanism, it can achieve a longer drive time with lighter weight than the active type while providing the same performance. Moreover, the proposed suit can control assistive power and timing using the spring force as the baseline output.

For assistive control, it is important to generate the assistive timing according to the movements of caregivers using our power-assisted suit. Various assistive timing generation methods have been proposed. A method using myoelectric potential has been established. However, it is difficult to implement because it requires installing and wearing sensors on the skin surface [5, 13, 14]. A method for inputting the timing by a wearer activating a switch has been put to practical use in some fields; agriculture, logistics, and construction field, however it is impractical for nursing care work because it is not hands-free [15].

In this study, we propose an assistive timing estimation method based on solely on the acceleration of the upper body. In this method, assistive timing is automatically generated by using internal sensors alone. The validity and effectiveness of the estimation results are evaluated based on experiments involving simulated nursing care motion.

2. Power-Assisted Lumbar Suit

The proposed power-assisted lumbar suit is based on the chest protector SK-696 manufactured by Komine Co., Ltd., in consideration of wearing comfort and ease. **Fig. 1** illustrates a 3D model of the assist suit. Six metal backbone plates are present on the back side of the chest protector, which are used to install a motor and springs. The backbone plates provide tensile force support to reduce the vertical load on the spine and to enhance the moment arm when the torso is extended. In addition, the backbone plate and protector limit the angle of the spine to prevent excessive assisting force from being applied to a human body. The total weight of the power-assisted suit is approximately 6 kg, which conforms to Japanese Industrial Standards [16].

The suit outputs assistive power using an electric motor and tension springs. Four springs are installed on

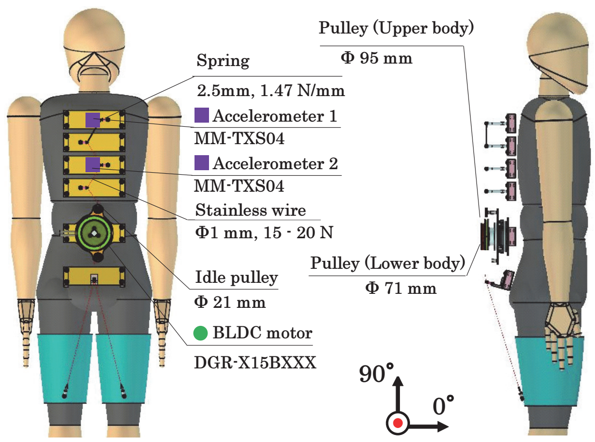


Fig. 1. Configuration of power-assisted lumbar suit.

(a) Back view (b) Side view

Fig. 2. Appearance of developed assist suit.

Table 1. Specifications of brushless DC motor.

Rated voltage	22 V
Rated load current	4.3 A
Rated output	59.4 W
Rated load	8 Nm
Mass	1000 g

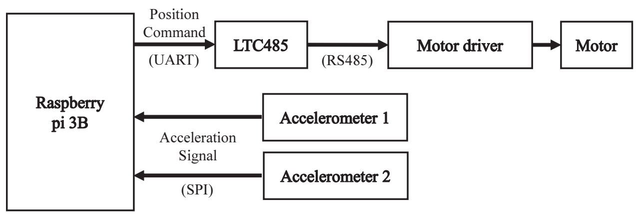


Fig. 3. Control system.

the upper part to extend the backbone, and two springs are installed on the lower part to assist the upper limbs during lifting motions. The tensile force of the spring is controlled by reeling in or relaxing the wires using the motor. A three-phase brushless DC motor (DRG-X15BXXX, Shinano Kenshi Co., Ltd.) is mounted at the center of the suit and reels in or out the upper and lower wires in tandem. The specifications of the motor are listed in **Table 1**.

To measure the motions of the wearer, two accelerometers (MM-TXS04) are mounted on a backbone plate, as depicted in **Fig. 1**. The measurement range of each accelerometer is ± 2 G. The pitch angles of the backbone plate θ_1 and θ_2 are calculated from each accelerometer as a coordinate, as depicted in **Fig. 1**.

2.1. Developed Power-Assisted Suit

Figure 2 illustrates the appearance of the developed power-assisted lumbar suit. The electric motor, tension springs, and stainless wires are mounted on the backbone plates. The backbone plates are covered by a metal mesh plate to prevent the wires from becoming entangled. All electrical components such as the computer, battery, and motor driver are installed on the side of the waist. The weight of the suit is approximately 6.0 kg with the battery and 5.2 kg without.

2.2. Control System

Figure 3 depicts the hardware configuration of the control system. The attitude angle of the upper body is calculated from the accelerations measured by accelerometers 1 and 2. The Raspberry Pi 3B computer outputs a rotation angle command to the motor driver according to the attitude angle. The motor driver communicates with the Raspberry Pi 3B by (RS485 communication) via am interface integrated circuit (LTC485). The motor is position controlled because the assistive torque can be calculated from the displacement of the springs, which depends on the reeled length of the wires.

3. Simulation of Digital Human Model

To determine a suitable magnitude of the assistive torque, we developed a digital human model using MATLAB/Simulink as shown in **Fig. 4**. The required assistive torque is estimated from the difference in joint torque with and without the suit, and a guideline for preventive measures against lower back pain is applied. The digital human model was designed for an adult male 170 cm tall in height, weighting 65 kg. The model consisted of 13 rigid body parts: head, body trunk, pelvis, upper and lower arm portions, femur, lower legs, and feet, which were connected using 12 joints: neck, shoulders, elbows, waist, hip, knees, and ankles [17]. The size and weight of each body part were determined with reference to the actual structure of the human body [18]. The Newton-Euler method and Featherstone's forward dynamics calculation method were used to perform the kinetic simulations. Input angles of joints of the human model were measured using the motion capture system OptiTrack (Natural Point Inc.), and the drive torque of each joint was calculated based on these measurements.

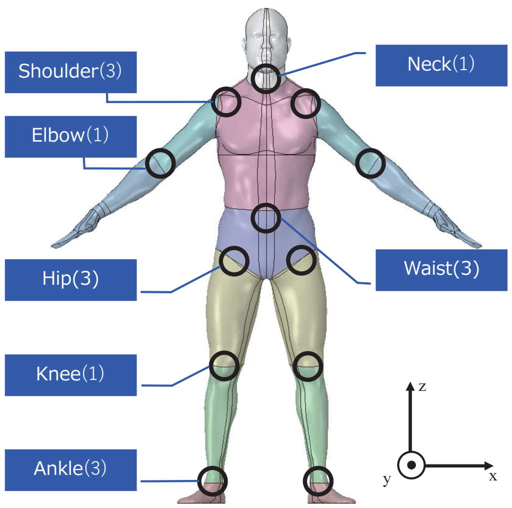
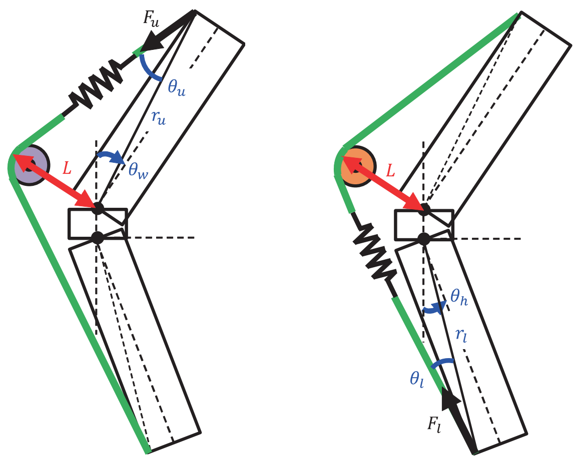


Fig. 4. Digital human model.



(a) Upper part (b) Lower part

Fig. 5. Simulation model (side view).

3.1. Assistive Torque

Figure 5 shows models of the power-assisted lumbar suit with upper and lower assistive parts. Assistive torques generated by the upper and lower parts respectively are calculated using Eqs. (1) and (2)

$$\tau_u = F_u r_u \sin\theta_u \quad \text{. . . . .} \quad (1)$$

$$\tau_l = F_l r_l \sin\theta_l \quad \text{. . . . .} \quad (2)$$

where, $r_u \sin\theta_u$ and $r_l \sin\theta_l$ are the moment arms of the upper body and upper leg, respectively. F_u and F_l are the assistive forces for the upper and lower parts respectively [11]. θ_w and θ_h are the pitch angles of the waist and hip, respectively, and L is the distance between the suit and the pulley.

The assistive force F_u in Eq. (1) has nonlinear characteristics with respect to the reeled wire length because the upper springs are installed alternately on the left and right. Therefore, the elastic characteristics of the upper spring were measured using a spring scale, and the results are shown in **Table 2**. In contrast, the assistive force of the lower part F_l changes almost linearly with the displacement of the springs. The spring constant of the lower part is 13.80 N/mm.

Table 2. Tensional force of upper spring unit.

Displacement [mm]	Weight [kgf]	Force [N]
0	0.00	0.00
10	0.30	2.94
20	0.45	4.41
30	0.75	7.35
40	1.10	10.78
50	2.00	19.60

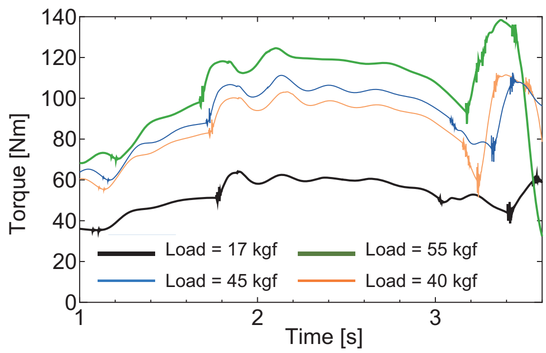


Fig. 6. Simulated torque.

Table 3. Reel wire lengths based on different load weights.

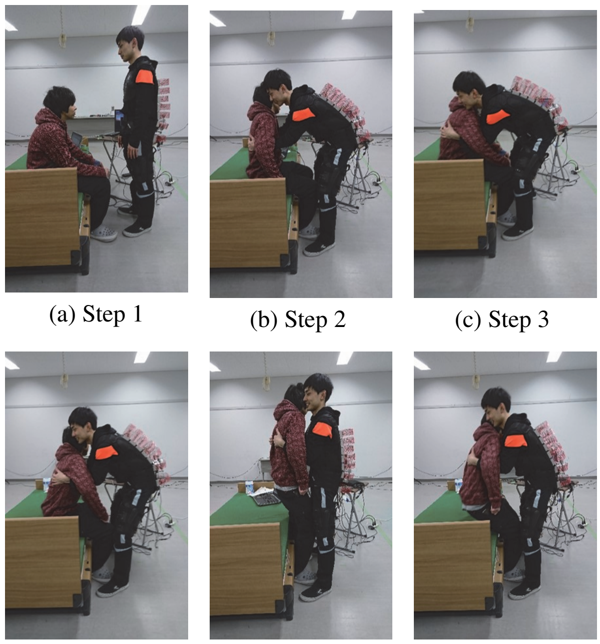
Load weight [kgf]	40	45	55
Required torque [Nm]	35	45	60
Upper reel [mm]	1	10	40
Lower reel [mm]	1	10	30

3.2. Simulation Result

To determine the magnitude of the assistive torque, load weights of 40, 45, and 55 kgf were simulated. These loads were determined based on the average body weight of elderly persons 75 years old or older; average male weight: 54.9 kg and average female weight: 42.2 kg. Here, the average body weight of the caregiver was 45 kg [19].

Figure 6 shows the simulated torque of the waist when the human model lifts loads of 17, 40, 45, 55 kgf. Here, 17 kgf is the maximum allowable load for women in their 40s, who make up a certain percentage of caregivers. The required assistive torque is a value that reduces the load on the wearer to the allowable load.

Therefore, the maximum reeling length of the wires was calculated according to the assumed loads: 40, 45, 55 kgf, as shown in **Table 3**. Here, the exerted torque of the suit is calculated by considering not only the wire lengths but also the extension of the springs due to the bending of the human body.



(a) Step 1 (b) Step 2 (c) Step 3

(d) Step 4 (e) Step 5 (f) Step 6

Fig. 7. Lifting procedure evaluated in experiments.

4. Lifting Motion

The lifting motion, which is a fundamental nursing care motion, places a heavy burden on the caregiver's lower back. Each experiment follows the lifting motion procedure shown in **Figs. 7(a)–(f)**. In **Fig. 7**, the caregiver holds the care receiver (step 2) and lifts him to a standing position (steps 3–5). After that, the care receiver is lowered to the bed gently (steps 6 and 1).

5. Assistive Timing Estimation

Assistive timing is as important as the assistive force to provide proper patient assistance. Therefore, assistive timing was determined by considering the experimental results of the lifting procedure based on three persons acting as caregivers and two persons acting as care receivers. The physical information and role of each participant are shown in **Table 4**. In the experiments, caregiver A lifted care receiver D, and caregivers B and C lifted care receiver E.

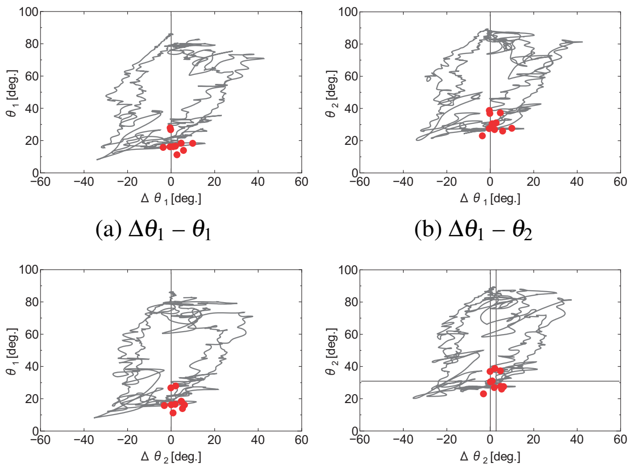
The caregivers controlled the start and stop of the assistive motion by controlling a switch during the lifting procedure, in which the motor reels in the wire while the switch is pushed. The upper-body acceleration and the assistive timing were measured several times for caregivers A, B, and C during the procedure.

Figures 8 and **9** show the relationships between the attitude angles of the spine (θ_1 or θ_2) and angle differences from 500 ms ago ($\Delta\theta_1$ or $\Delta\theta_2$). Here, the angles θ_1 and θ_2 correspond to accelerometers 1 and 2 shown in **Fig. 1**, respectively. The push and release timings of the switch are plotted in **Figs. 8** and **9**.

Table 4. Physical information of participants in lifting experiments.

Subject	Role*1	Sex	Age	Height [cm]	Weight [kgf]
A	G	Male	24	169	60
B	G	Male	22	172	65
C	G	Male	22	165	60
D	R	Male	24	171	68
E	R	Male	22	158	53

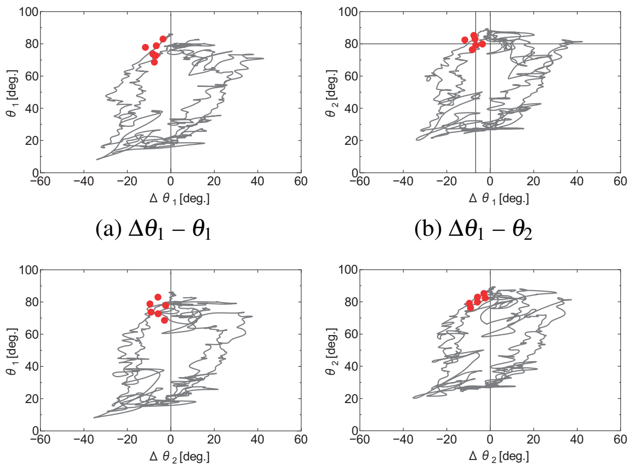
*1G: caregiver; R: care receiver.



(a) $\Delta\theta_1 - \theta_1$ (b) $\Delta\theta_1 - \theta_2$

(c) $\Delta\theta_2 - \theta_1$ (d) $\Delta\theta_2 - \theta_2$

Fig. 8. Assistive timing (switch is "on").



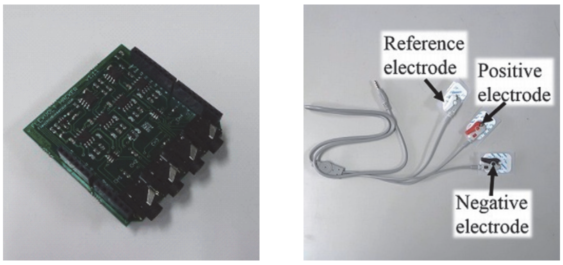
(a) $\Delta\theta_1 - \theta_1$ (b) $\Delta\theta_1 - \theta_2$

(c) $\Delta\theta_2 - \theta_1$ (d) $\Delta\theta_2 - \theta_2$

Fig. 9. Assistive timing (switch is "off").

The relationship between the attitude angle and angle difference creates a closed cycle. The angle difference takes negative values during flexing, and takes positive values during extension.

Figures 8(d) and **9(b)** have the lowest variance in plotted timing compared to the others. Therefore, the criteria for assistive and release timings are formulated using



(a) Amplifier board (b) Myoelectric pads

Fig. 10. Myoelectricity sensor.

Eqs. (3) and (4), respectively. When the condition indicated by Eq. (3) is satisfied, the motor rotates to 50° to provide the assistive force by tightening the wires. In contrast, when the condition indicated by Eq. (4) is satisfied, the motor rotates to −80° to release the wires. The setup is designed so that when the rotation angle of the motor is 0°, which corresponds to the wearer having an upright posture, and some tension is applied to the springs to prevent that the wires do not sag.

$$\theta_2 \geq 31.3^\circ \ \cap \ \Delta\theta_2 \geq 2.9^\circ \quad (3)$$

$$\theta_2 \leq 80.9^\circ \ \cap \ \Delta\theta_1 \leq -7.4^\circ \quad (4)$$

6. Experiments

The performance of the proposed assist suit was evaluated by comparing the experimental results obtained with and without the developed suit. The assistive force exerted by the suit was evaluated based on the muscle surface myoelectric potential.

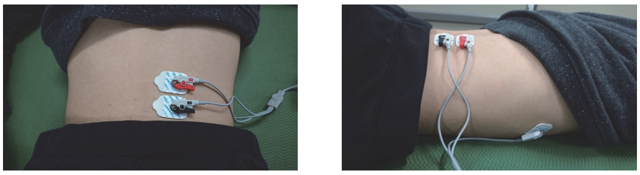
6.1. Myoelectricity Sensor

We evaluated the effectiveness of the power-assisted suit based on the myoelectricity of the erector spinae muscles, which is a muscle group used to straighten the body.

The myoelectricity was measured using an electromyography (EMG) sensor (ArduinoEMGShield), as shown in **Fig. 10(a)**. The gain of the amplifier was 1845, and the sampling frequency was 2 kHz.

In the measurement of the surface myoelectric potential, noise will occur according to the skin surface condition. Therefore, the measurement area was cleaned with alcohol to remove any contaminants before measurement. **Fig. 10(b)** shows the attached myoelectric pads. The anode and cathode electrodes are attached to the surface of the erector spinae muscle, and a reference electrode is attached to a costal bone that is not covered by muscles, as shown in **Fig. 11**.

In this study, the effectiveness of the assist suit is evaluated from the quantitative measurement of the surface myoelectric potential as muscle activity [20]. The maximum amplitude of the moving root mean square value (moving RMS) for 300 ms was employed for evaluation because the myoelectric potential is an AC signal.



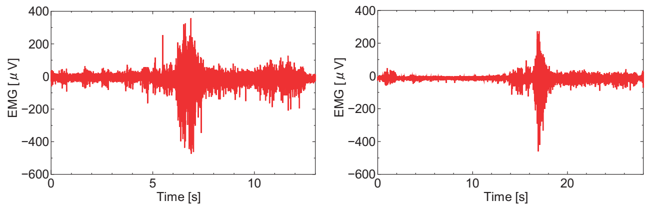
(a) Anode, cathode (b) Reference

Fig. 11. Positions of myoelectric pads

Table 5. Physical information of participants in lifting experiments.

Subject	Role*[1]	Sex	Age	Height [cm]	Weight [kgf]
A	G	Male	24	169	60
F	G	Male	23	175	60
G	G	Male	23	161	58
D	R	Male	24	171	68
E	R	Male	22	158	53

*[1] G: caregiver, R: care receiver.



(a) Without suit (b) With suit

Fig. 12. Surface myoelectric potential of subject A.

6.2. Experimental Conditions

For the suit evaluation experiment, the EMG experienced by caregivers with and without the suit were compared based on the simulated nursing care lifting procedure shown in **Fig. 7**. In the experiment, the lifting motion was measured for three persons acting as caregivers, and the physical information of the caregivers and the care receivers are shown in **Table 5**. Subjects A, D, and E represent the same persons as in the timing determination experiment, whereas subject F and G are new subjects. Caregiver A lifts care receiver D, and caregivers F and G lift care receiver E.

6.3. Experimental Results

Figures 12(a) and **(b)** show the surface myoelectric potential of the erector spinae muscle of subject A without and with the suit. From **Fig. 12**, it was confirmed that the myoelectric potential level decreased when the suit was worn. **Figs. 13(a)**, **(b)**, and **(c)** show the moving RMS of the surface myoelectric potential of subjects A, F, and G, respectively, used to quantify the load. It was confirmed that the myoelectric potential level decreased for most of the movements performed by all three caregivers. The decrease in the peak load is remarkable, whereas the change

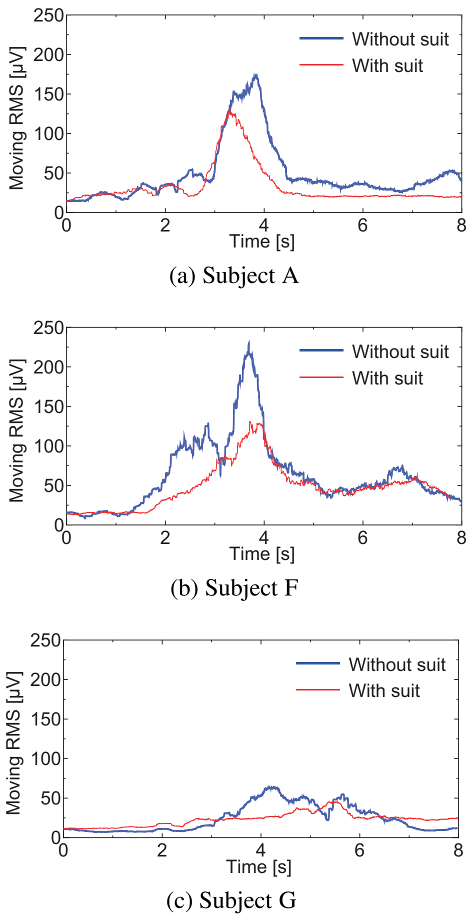


Fig. 13. Moving RMS of the surface EMG.

Table 6. Maximum of moving RMS and improvement rate.

Subject	Without suit [μV]	With suit [μV]	Improvement rate [%]
A	174.1	113.2	35.0
F	175.5	148.2	15.6
G	86.2	43.3	49.8

in the peak for lighter loads is small. This indicates that the suit exerted an assisting force at the correct timing.

Table 6 shows the peak of the moving RMS value and the improvement rate for each subject. Each value in **Table 6** represents the average of three experiments conducted for each caregiver/receiver pair. From **Table 6**, the maximum value of the moving RMS decreased when the lumbar assist suit was worn by all caregivers.

In the measurement experiments, caregiver A reported an insufficient amount of wire reeling. This occurred because the assistive force of the suit was designed by assuming that middle-aged women care for last-stage elderly patients. Therefore, it is considered that the assistive force was insufficient in this experiments by adult males. Subject F reported a timing mismatch between the lifting motion and wire reeling. From **Fig. 13(b)**, although the assist was performed properly, the subject felt some discomfort because of an inconsistency in assistive timing resulting from individual physical differences among caregivers.

Moreover, **Table 6** shows that the improvement rate varies widely depending on the caregiver, and a negative relationship was observed between the subject's height and the improvement rate. Here, caregiver F was tallest, caregiver A was next tallest, and caregiver G was shortest. Further research on the amount of assistive force required according to different physical traits is required.

7. Conclusion

In this paper, we discuss the assistive torque and assistive timing of the proposed lumbar power-assisted suit. The assistive torque is determined based on a simple digital human model, and assistive timing is estimated from the acceleration of upper body of the wearer. The proposed control method was evaluated based on the lifting procedure common to nursing care based on three persons acting as caregivers and two persons acting as care receivers. The myoelectric potential on the surface of the erector spinae muscles during lifting with and without the suit was measured and compared. From the experimental results, it was confirmed that the lifting motion was assisted by the suit, although one caregiver felt discomfort because of an inconsistency in the assistive timing. Adjusting the assistive timing according to individual differences and quantitative verification of the assistive force are future tasks. In addition, experiments in actual nursing care sites or more detailed simulated environments are required to verify the potential of the power-assisted suit for practical applications.

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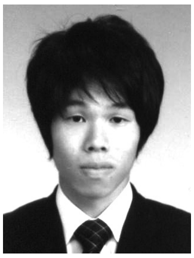

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- "Study on Output Prediction System of Wind Power Generation Using Complex-valued Neural Network with Multipoint GPV Data," IEEJ Trans. on Electrical and Electronic Engineering, Vol.8, No.1, pp. 33-39, Jan. 2013.

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- "Multiple Vehicles Semi-Self-driving System Using GNSS Coordinate Tracking under Relative Position with Correction Algorithm," Int. J. of Advanced Computer Science and Applications (IJACSA), Vol.8, Issue 8, pp. 418-426, 2017.

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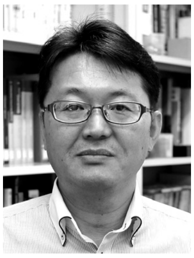

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- "Human-Friendly Safe Driving Support System of Electric Wheelchair for Drift Suppression," J. of Signal Processing, Vol.24, No.4, pp. 199-202, 2020.

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