2025 IEEE/SICE International Symposium on System Integration (SII)
January 21-24, 2025, Munich, Germany

# Design and Verification of Torque Pattern for Reducing Mental Load in Artificial Muscle Assist Devices

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*Abstract*— **Seventeen percent of the Japanese workforce performs physical labor, and there is a need to reduce the physical burden of this work. Assist devices using artificial muscles, which are lightweight and easy to install, are effective in reducing the burden of physical labor. However, when an assist device is used for lifting a load, for example, the assist torque must be quickly raised to a certain level, which may lead to discomfort for the user, reduced operability. Therefore, this research conducted an experiment to verify whether controlling the assist torque of an assist device using artificial muscles according to the previously measured exerted muscle strength pattern of the lifting motion can reduce the mental load compared to a case where the assist torque is suddenly raised. Participants in the experiment were asked to receive assistance from the assist device, and the results were evaluated based on subjective evaluation and the sum of squares of the joint angular jerk, an index of discomfort. The results showed that there was no significant difference in subjective mental load, but the sum of squares of joint angular jerk decreased significantly.**

## I. INTRODUCTION

The percentage of workers engaged in physical work in Japan is 17% [1], and back pain and physical fatigue among physical workers have become a problem. assist devices can be used to reduce the physical burden of physical work. So far, assist devices that can be used for physical work have been developed, such as HAL (Cyberdyne Inc.) [2] which uses a motor, and DARWING Hakobelude (Daiya Kogyo Inc.) [3] which uses pneumatic artificial muscles. Pneumatic artificial muscles are suitable actuators for assisting physical work that involves repeated lifting of heavy objects for long periods of time because of their low rigidity, high safety, and high exerted force per mass. In this research, we focused on assist devices using pneumatic artificial muscles.

One of the challenges in using assist devices is the existence of mental load, such as difficulty in handling the device when using it [4]. Mental load may cause a decrease in the ability to concentrate on work, so it is necessary to prevent the occurrence of mental load. To date, efforts have been made to reduce mental load by matching the timing of the assistance to the motion, including the development of a device that controls the assistance timing by combining information on EMG and joint angles [5]. In addition, the force pattern of the force to raise the assist torque from zero to a certain level may also be related to the mental load. It has been reported that people exert maximum joint torque at the timing when the load leaves the ground in the lifting motion of a load, which is often performed in physical labor. [6]. Therefore in the operation of lifting a load, and to effectively support the lifting operation, it is necessary to strongly raise the assist torque to the maximum torque at the moment the lifting operation starts. Such a sudden change in assist torque may cause a mental load. According to a research that evaluated the mental load in human-machine contact motions, a strong discomfort is reported to occur when a large force is suddenly applied to a person [7], and an assist device may cause discomfort in the same way. In addition, it has been shown that high acceleration of the body caused by a sudden assistance decreases the operability of assist devices [8]. Therefore, it is necessary to control the assist torque in such a way that the assist torque can be raised in a short period of time while at the same time preventing the generation of mental load. Ikeura et al. have succeeded in reducing the mental load of the users by admittance control of the assist force based on the user's operation force in a stationary assist device [9]. However, this system requires a force sensor to be attached to the raised object, making it difficult to apply this control method to a wearable assist device. To solve this problem, It is necessary to estimate the muscle exertion pattern of a person when lifting a load in advance, and to make the assist device exert its assist torque accordingly, so that the assist torque is in harmony with the person's muscle exertion.

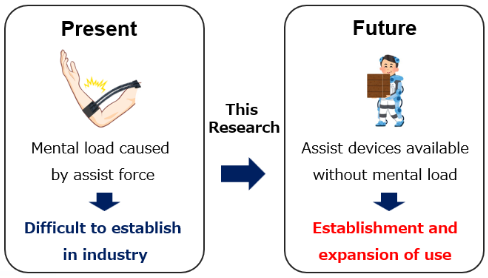


Fig. 1. Research objective

The purpose of this research was to verify the effect of the assist torque on reducing the mental load of the user, which is a fixed percentage of the muscle strength exerted by the user. Fig.1 shows the objective of this research.

*This work was supported by JSPS KAKENHI Grant Numbers 22K18418.

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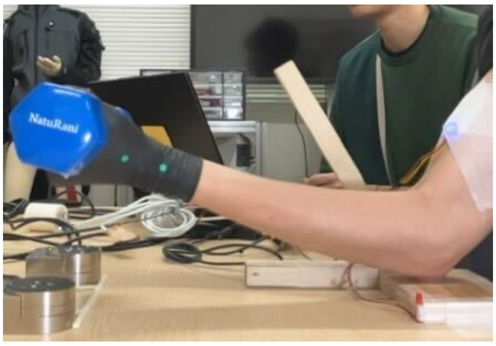

Fig. 2. Exerted muscle strength estimation experiment

## II. EXERTED MUSCLE STRENGTH ESTIMATION EXPERIMENT

An experiment was conducted to estimate the muscle strength exertion pattern of a person in order to design an assist torque that is a fixed percentage of the exerted muscle strength when a person lifts a load. This experiment was conducted with the approval of the Ethics Committee of the Graduate School of Advanced Science and Engineering, Hiroshima University (ASE-2023-24).

### A. Method

Ten healthy male participants, aged 22-23 years, participated in the experiment. Fig.2 shows the experimental situation. First, the participants sat on a chair facing forward. Then, gloves were worn and markers were affixed to the surface of the gloves at two locations: at the center of the wrist joint and at the side of the hand. Then, the participant placed his elbow on a pedestal and grasped a 3 [kg] dumbbell. The participants then performed the task of lifting the dumbbell by bending the elbow joint 60 [deg] in accordance with the angle indicator placed near the participant. An electronic metronome was used to unify the rhythm of each participant's motion during the task. The interval between tones was set to 0.8 [s], and the participants were instructed to complete the lifting motion between two tones. Five trials of the task were performed for each participant.

To estimate elbow joint torque during the dumbbell ground phase, two force sensors (PFS055YA501U6IO, Leptrino, Inc.) were placed under each end of the dumbbell to measure the load from the dumbbell at 1200 [Hz]. The motion of the elbow and wrist joints was also measured to estimate the elbow joint torque during lifting. A camera (iPhone12, Apple) was placed in front of the marker in the sagittal plane of the participant, and the attached marker was captured on video at 240 [Hz]. In addition, the electromyography (EMG) of the biceps brachii muscle were acquired in order to clarify the timing between muscle strength exertion and EMG elevation, so that the EMG can be utilized as triggers for the assist device. A EMG sensor (ALTs Inc., Myoscan) was attached to near the biceps brachii muscle and measured at 500 [Hz].

### B. Data analysis

From the obtained data, elbow joint torques were obtained using different analysis methods during the pre-lifting phase when the dumbbells were on the force sensors and during the lifting phase when the dumbbells were off the force sensors. First, elbow joint torque was obtained from the magnitude of the force applied to the dumbbell during the pre-lifting phase. The time series data of the force applied by the subject to the dumbbell, $f$, can be expressed as follows.

$$f = m_d g - f_s \tag{1}$$

where $m_d$ is the mass of the dumbbell, $g$ is the acceleration of gravity, and $f_s$ is the sum of the output values of the two force sensors. The time series data $\tau_g$ of elbow joint torque is obtained using $f$ as follows

$$\tau_g = fr + m_h g l \tag{2}$$

where $r$ is the distance from the elbow joint center to the dumbbell grasping position, $m_h$ is the mass of the hand and forearm, and $l$ is the distance from the elbow joint center to the center of gravity of the hand and forearm.

The elbow joint torque during lifting was obtained from the equations of motion of the elbow and wrist joints.The elbow joint and wrist joint angles were obtained from the data of markers attached to the participant. First, the pixel coordinates of the markers were obtained from the captured data using OpenCV, an image processing library in Python. Then, the pixel coordinates were processed with a low-pass filter (cutoff frequency: 5 [Hz]), and the elbow and hand joint angles were calculated using the inverse tangent function from the joint centers and the pixel coordinates of the markers.Then, by substituting the angle, angular velocity, and angular acceleration of the elbow and wrist joints into the equations of motion, time series data of the elbow joint torque $\tau_a$ can be obtained as follows.

$$\tau_a = \boldsymbol{M}(\theta)\ddot{\boldsymbol{\theta}} + h(\theta, \dot{\theta}) + G(\theta) \tag{3}$$

where $\ddot{\boldsymbol{\theta}}$ is the angular acceleration vector, $\boldsymbol{M}$ is the inertia row vector, $h$ is the nonlinear term, and $G$ is the gravity term. Parameter values such as the mass and moment of inertia of the body segment in (1), (2) and (3) are obtained from height and weight data of participants and previous studies on body inertia coefficients of young Japanese [10].

The obtained EMG data were subjected to linear trend removal, full-wave rectification, and low-pass filtering (cutoff frequency: 5 [Hz]).

### C. Result

Fig.3 shows the results of the estimation of exerted muscle strength during dumbbell lifting and the waveform-processed EMG. The blue and red lines indicate the exerted muscle strength during dumbbell grounding and lifting, and the yellow line indicates the waveform-processed EMG. The semi-transparent fill in each color indicates the 95% confidence interval. The results show that the participant gradually raises

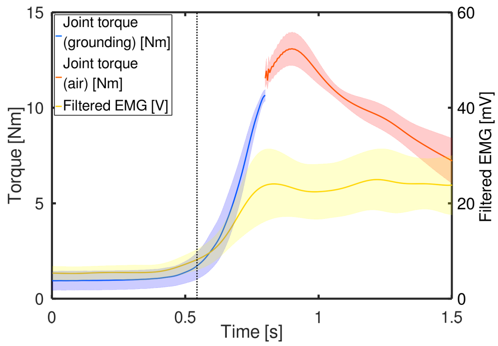


Fig. 3. Exerted muscle strength and filtered EMG in dumbbell lifting task

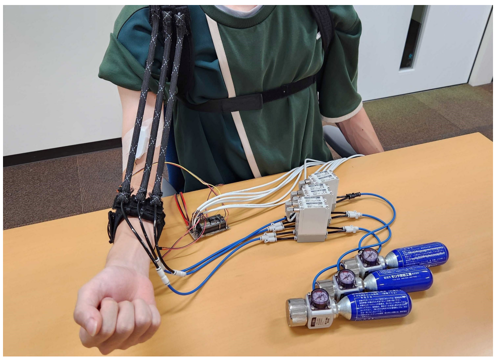

Fig. 4. Elbow joint flexion motion assist device

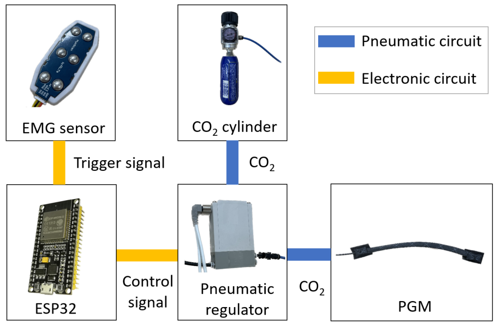


Fig. 5. System configuration of assist device

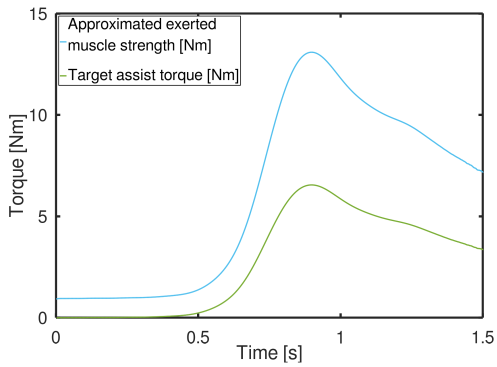


Fig. 6. Approximated exerted muscle strength and target assist torque

muscle strength over a period of about 0.5 [s] in the dumbbell lifting task.

The EMG analysis results show that the rise in EMG and the exertion of muscle strength started at approximately the same time. The time when the EMG raised to 1.5 times the resting level is indicated by the black dotted line in Fig.3, indicating that the start of muscle exertion can be detected at an early stage based on EMG. Therefore, it is possible to use EMG as a trigger for the start of assist.

## III. ELBOW JOINT FLEXION MOTION ASSIST DEVICE

### A. Device Configuration

In this research, a assist device to support elbow flexion motion was used, and an overview of the used device is shown in Fig.4. This device utilizes three low-pressure-driven pneumatic gel muscles (PGMs) [11] as actuators for assist. The device consists of a PGM, a $CO_2$ cylinder, a pneumatic regulator (SMC Inc., ITV0031) that supplies and exhausts the PGM, a EMG sensor (ALTs Inc., Myoscan), a microcomputer (Espressif Systems Inc., ESP32) for force control of the PGM, and supporters to secure the PGM to the body.

### B. Assist torque control

Fig.5 shows the configuration of the force control system of the developed device. The pneumatic regulator is connected to a $CO_2$ cylinder with the supply pressure set to 0.2 [MPa] on the input side and a PGM on the output side. The pneumatic regulator proportionally controls the pressure supplied to the PGM based on the input voltage signal from the ESP32. The trigger for starting the assistance was a EMG signal; the potential of the biceps brachii muscle was measured at 500 [Hz], and after waveform processing (linear trend removal, full-wave rectification, and low-pass filtering (cutoff frequency: 5 [Hz])), the assistance was started when the EMG exceeded a threshold value. Based on the results of the exerted muscle strength estimation experiment, the threshold value was set to 1.5 times the EMG after waveform processing in the 0.5 [s] resting state before the start of the assist. The pattern of the assist torque was designed to be 50% of the exerted muscle strength during the muscle strength estimation experiment, as shown in the green line in Fig.6. First, In order to obtain a waveform that is continuous and closely approximates the experimentally obtained torque waveform, the exerted muscle force data was approximated by a 50th order polynomial equation (light blue line). The target assist torque was designed to be 50% of the exerted muscle strength by eliminating the offset value before the exertion of muscle strength.

Fig.7 shows the assist torque waveform (Type A), which is 50% of a person's exerted muscle strength realized by the assist device. Fig.8 shows the assist torque waveform when the input voltage is raised in a stepwise manner (Type B). The blue line indicates the assist torque. The red line shows the input voltage of the pneumatic regulator, and the assist torque is designed to approximate the target assist torque by adjusting the input voltage rise. The green line in Fig.7

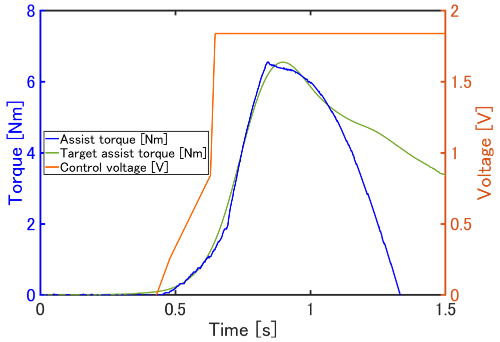


Fig. 7. Assist torque as 50% of exerted muscle strength (Type A)

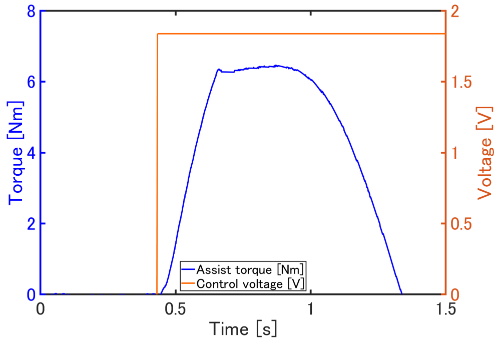


Fig. 8. Assist torque when step-like input voltage is applied (Type B)

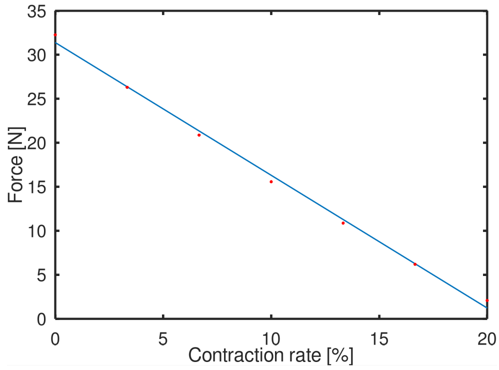


Fig. 9. Contraction rate-force characteristics for a PGM

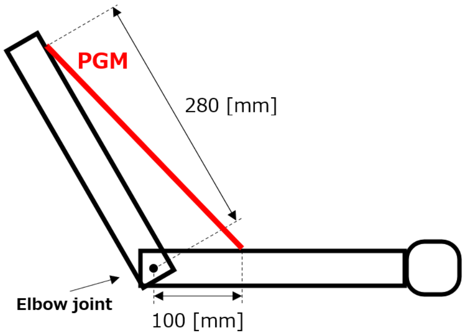


Fig. 10. Position of the PGM on the body

shows the target assist torque shown in Fig.6. The assist torque shown in Fig.7 and Fig.8 is calculated by considering the contraction rate-force characteristics of the PGM shown in Fig.9. First, we measured the contraction force of the PGM when it was in its natural length. Then, based on the model of the position of the PGM attached to the person shown in Fig.10 and the elbow joint angle during the exerted muscle strength estimation experiment, we calculated the time-series change in contraction rate of the PGM. Then, we calculated the contraction force that takes into account the contraction rate of the PGM. Finally, the assist torque was calculated from the model shown in Fig.10 based on the contraction force of the PGM. In this experiment, the PGM begins to contract before the dumbbell leaves the force sensors because the start of muscle exertion is detected by EMG data. The length of the PGM did not change significantly before the dumbbell leaves the force sensors (before 0.8 [s] in Fig.7). On the other hand, after the dumbbell leaves the force sensors, the PGM shortens and the assist torque applied to the user decreases (after 0.8 [s] in Fig.7). When the elbow joint flexes 30 [deg], the contraction rate of the PGM reaches 20% and the PGM can no longer provide assist torque. Therefore, in the latter half of the assist, the assist torque is much lower than the target assist torque, but the waveform is well approximated during the phase when the assist torque is raised. Type A is characterized by a more gradual rise in torque, especially in the first half of the rising, compared to the assist torque waveform of Type B.

## IV. EXPERIMENT TO EVALUATE DESIGNED ASSIST TORQUE

### A. Method

An experiment was conducted to verify the effect of the assist torque, which was set at 50% of the exerted muscle strength, on the reduction of mental load. 10 healthy male participants aged 22-24 years cooperated in the experiment. As shown in Fig.4 and Fig.10, the participants wore a supporter on the shoulder and forearm, and the assist device with both ends of the PGM fixed to the supporter. Then, EMG sensors were attached to the area corresponding to the biceps brachii muscle. Then, the participant lifted a dumbbell while receiving the assist, using the same procedure as that used in the exerted muscle strength estimation experiment. Assist torque was set to three conditions: Type A, Type B, and Type C (unassisted condition). Type B is a method that has been used in previous studies of various assist devices [12]. The Type C test was conducted to verify how much the mental load was reduced by the Type A assist torque compared to the unassisted case.

After the completion of the task, a subjective evaluation of mental load was conducted using the NASA-TLX [12] (a subjective evaluation index consisting of six scale items: mental demand, physical demand, temporal demand, performance, effort, and frustration. VAS evaluation with values ranging from 0 to 100, the higher the rating value, the higher the mental load.) Questions were asked about mental demand "How much mental and perceptual activity was required?", about the physical demand "How much physical activity was required?", about the temporal demand "How much time pressure did you feel on the speed of the task?", about the performance "How successful were you in achieving the goal of the task?", about the effort "How hard you worked mentally and physically?", and about the frustration "How anxious, irritated, stressed, or discomfort did you feel?".The experiment employed an discomfort and task-intensive task in which the subject had to raise a dumbbell to a predetermined position over a period of time while receiving strong assistance. Therefore, we hypothesized that changes in the manner in which the assist torque is applied to the subject in particular would be expressed in the evaluation value of frustration, and that changes in operability due to the assist torque pattern would be expressed in the evaluation value of mental demand, temporal demand, and performance. The task was performed in a total of three sessions, with five trials per condition of assisting force and a five-minute break after five random trials in one session.

In order to evaluate the mental load from the joint motion, the elbow joint angle was measured using the same procedure as in the exerted muscle strength estimation experiment.

### B. Data analysis

The mean and standard deviation of the NASA-TLX evaluation values were calculated for each evaluation item for all participants on all trials. Also, as an indicator of discomfort, one of the mental load, the sum of squares of the elbow joint angular jerk during dumbbell lifting was calculated. According to a previous research, the body angular jerk when receiving force from an assist device is related to discomfort [14], [15] and can be used as an objective index of discomfort. The sum of squares of the angular jerk $S$ is expressed as follows.

$$S = \sum_{k=1}^{n} {J_k}^2 \tag{4}$$

where $J_k$ is the angular jerk of the elbow joint, $n$ is the number of samples during the time from the start of assistance until the elbow joint angle reaches 30 [deg], and the sum of squares of angular jerk is calculated during the time when the participant can receive assistance. The higher the $S$, the greater the discomfort.

Significant difference tests were conducted on the sum of squares of angular jerk and NASA-TLX for the three conditions of the assist torque pattern. For each evaluation item, the null hypothesis was "No difference in subjective evaluation and sum of squares of angular jerk for the three conditions" and Significant differences were tested by the Friedman test at a significance level of 5%. For the endpoints for which significant differences were found, Wilcoxon's signed rank test (Holm-Bonferroni correction) was used to test for significant differences in each assist torque condition at the 5% level of significance as a post-hoc test. (*: $p < 0.05$, **: $p < 0.005$)

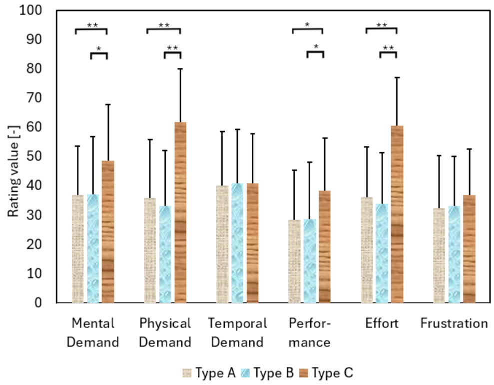


Fig. 11. NASA-TLX rating values for each assist torque condition

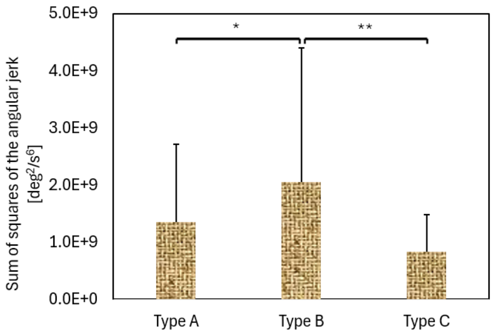


Fig. 12. Sum of squares of elbow joint angular jerk for each assist torque condition

### C. Result and discussion

Fig.11 shows the NASA-TLX evaluation values for each assist torque condition. As a result, there was no significant difference in the evaluation values among the different assist torque raising patterns for all evaluation items. This result contradicts the expectation that Type A would have lower rating value of mental demand and frustration than Type B. However, in mental demand, physical demand, and effort, Type A and Type B had significantly lower rating value than Type C. This indicates that the subjective lightening of work was more successful than in the case of no assistance. The performance was also significantly lower for Type A and Type B. This suggests that the assisted task was easier to accomplish within the time limit.

Fig.12 shows the sum of squares of the angular jerk of the elbow joint under each assist torque condition; the sum of squares of the angular jerk of Type A was significantly smaller than that of Type B. Here, since large angular jerk generates discomfort, we focus on the highest and lowest values of angular jerk. Fig.13 shows the angular jerk changes during assistance. The time of the highest value of angular jerk corresponds to the phase in which the assist force is rising, and the time of the lowest value corresponds to the phase after the assist torque has fully risen. The reason for the maximum value is that Type A assisted the participant gradually in a pattern similar to that of human exerted muscle strength, so that a strong force was not suddenly applied to

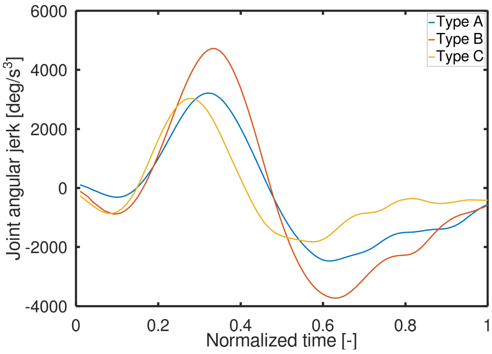


Fig. 13. Elbow joint angular jerk during assistance for each assist torque condition

the participant, and the reason for the lowest value is assumed to be that the participant suddenly relaxed after the force was suddenly applied in Type B. There was no significant difference in the sum of squares of the angular jerk between Type C and Type A. It is known that humans minimize their body dynamism in the task of moving their hand to the target position [16], and the sum of squares of the angular jerk of Type C, which did not receive any assistance, was the smallest. Type A's angular jerk degree waveform is close to that of Type C. It can be said that Type A is able to assist without significantly interfering with natural human joint motion.

In this experiment, there was no significant difference in the subjective evaluation index, although there was a significant difference in the motion index of discomfort between the difference in the assist torque pattern. This may be due to the fact that the assist torque was not large enough to cause a large change in mental load, although the participant's joint motion changed with the difference in assist torque patterns. In addition, the task in this experiment required a high level of concentration on the task itself, such as matching the rhythm of the metronome and bending the joint angle to 60 [deg]. This may have made it difficult to be conscious of the assist torque itself, and thus made it difficult for differences in subjective evaluations to emerge.

## V. CONCLUSIONS

The purpose of this research was to develop an assist device using pneumatic artificial muscles, and to verify the effect of the assist torque on reducing mental load, which was set as a 50% of a person's exerted muscle strength in a task. As a result of experiment, the sum of squares of the elbow joint angular jerk, which is an index of discomfort, was significantly smaller in the case of the assist torque pattern in which the assist torque was a 50% of the person's exerted muscle strength than in the case in which the assist torque was applied rapidly by raising the assist torque, which has been commonly used. However, there was no significant difference in the subjective evaluation indices by the assist torque pattern. Therefore, the assist torque pattern designed in this research could not reduce the mental load other than discomfort, but may have reduced discomfort.

Future issues include increasing the ratio of assist torque to exerted muscle strength (assist ratio) and examining changes in mental load. In this research, the experiment was conducted with an assist ratio of 50%. However, if the assist ratio is further increased, the force applied to the user and the effect on joint motion will increase, and there is a possibility that significant differences in subjective evaluation indices will be observed depending on the torque pattern.

## ACKNOWLEDGMENT


This work was supported by JSPS KAKENHI Grant Numbers 22K18418.


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