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Validation of a lower limb exoskeleton assist device focusing on viscous properties: verification of assist effectiveness by measuring muscle activity

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Abstract

Because exoskeletal assistive devices are worn directly by a person, enhancing cooperation is important. However, existing assistive devices have problems in terms of their cooperation with human behaviors. This is because existing assistive devices are driven by estimating the wearer's movement intention based on predetermined movement time and device angle information. Although these methods are expected to work as expected, in practice, it is difficult to achieve the expected behavior. Therefore, an assistance method is required to reduce such misalignment with time and misalignment between the device and wearer. Therefore, this study focused on the viscoelastic properties that generate force in response to movement and are expected to enhance coordination. In a previous study, the authors confirmed the effects of viscoelastic properties or an assistive device with variable stiffness. However, viscous characteristics during movement have not been considered. In this study, we aimed to improve the coordination by focusing on the viscous characteristics. The viscous torque outputs in response to the angular velocity are expected to be driven in response to actual human motion. In this study, the viscous torque was calculated as the product of the command viscosity coefficient and the joint angular velocity and was applied to a lower-limb exoskeleton-type assist device equipped with a magneto-viscous fluid brake and a planetary gear mechanism. In addition, a viscous command that changes the torque according to speed (proposed method), a time command that changes the output value according to the passage of time, and an angle command that changes the command value according to the angle information of the device were applied to the assist device, and surface EMG measurements and command signals were compared. The target movement was a seated movement, and the left and right vastus medialis and semimembranosus muscles were measured. More than half of the subjects showed a decrease in myopotential for five subjects for all three command methods, confirming the effectiveness of the viscous command.

Keywords Exoskeletal assistive device, EMG, Viscous characteristics, Seated movement

Introduction

Because exoskeletal assistive devices are directly attached to a person, it is important that they cooperate with that person's behavior. Ideally, assistive devices should be driven voluntarily, similar to human muscles; however, it is difficult to read and drive human intentions accurately. Several existing assistive devices do not voluntarily decide to drive but rather output

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preprogrammed commands based on information obtained from the device's sensors or perform feedback control using device information. However, these methods suffer from problems with cooperation. In a method that outputs preprogrammed commands based on the sensor information of the device, it is difficult for the wearer to act as expected every time, and a misalignment exists between the expected and actual assistance. In particular, when reading the start of the operation from the sensor information and outputting commands according to the passage of time, it is desirable for the wearer to operate according to a predetermined time; however, it is difficult for the wearer to operate according to a set time. On the other hand, when feedback control is performed based on device information, it is difficult for the motion trajectory of the wearer and device to match perfectly, making it difficult to estimate the correct motion intention. In particular, the larger the output value of the device, the larger the discrepancy between the device and wearer. Such misalignments between the set time and actual movement time or between the motion trajectory of the device and the wearer can cause problems in terms of coordination, such as loss of assist force when assistance is required or force in a direction not anticipated by the wearer. Furthermore, it is difficult to completely eliminate such misalignments. Therefore, an assistive method with a wide tolerance for misalignment is required.

As a method to reduce misalignment, existing studies have included assistive orthoses for the wearer that consider changes in the center of rotation of the knee [1, 2]. This change can reduce the misalignment between the device and wearer because the center of rotation changes when the knee is in motion. However, the magnitude and change in misalignment when torque is applied to the device have not yet been verified. On the other hand, research has been conducted on the mechanism for automatic adjustment of misalignment [3]. This method studies the effects of misalignment under different misalignment conditions and at different levels of assist torque and finds that the pseudo-interaction force and torque are not affected by misalignment or the magnitude of the assist torque. However, even with this method, device misalignment is not completely eliminated. On the other hand, methods have been proposed to improve the overall human-device coordination. Examples include impedance controls [4] and variable viscoelastic joints [5]. However, impedance control cannot handle sudden inputs that exceed the actuator performance. The variable viscoelasticity parameters are identified using online parameter identification; however, because they are based on information from the device, the effect of misalignment between the device and the wearer is considered to be a factor. Therefore, an assistance method that can easily reduce misalignment is required.

Because viscoelasticity generates forces in response to actions, cooperative behavior is expected to be achieved even if the device is not voluntarily controlled. To enhance the cooperation with humans, studies have focused on the variable viscoelastic properties of humans [6]. Research on variable viscoelasticity has shown that viscoelasticity influences the assistance. An assist with variable stiffness that changes the stiffness during movement was also proposed, and the assist effect was confirmed [7]. In contrast, viscosity was considered as a variable viscoelastic element in a previous study, but it was the constant viscosity during motion and variable viscosity during operation were not considered. In this study, we aimed to assist the knee joints with variable viscosities during motion. In a previous study, it was confirmed that the command-timing deviation was reduced by a constant-viscosity assist [8]. However, the effect of viscous assistance has not been confirmed in previous studies.

In this study, the viscous torque obtained by the product of the angular velocity and the viscosity coefficient was applied to a lower limb exoskeleton assist device using a magnetorheological fluid brake (hereinafter, referred to as "MR brake") to experimentally verify the assist effect by measuring muscle activity. In this study, we focus on seated movement as the movement to be assisted by viscosity and calculate the command viscosity coefficient using motion analysis. Viscous assist provides assistance for passive movements, but we believe that passive movements at the knee joint, such as seated movements and stair descent, are important in daily life. The assist device was equipped with an MR brake that expressed the apparent viscosity, and a planetary gear mechanism that increased the output. Although MR brakes are used in this study, we believe that the proposed viscous assist method can be applied to other actuators such as motors and soft actuators. The muscle activity during seated movements was measured using an assistive device. To compare the viscous assist device, we also measured the amount of muscle activity for a time command, in which the command value changed with time, and for an angle command, in which the command value changed with the angle of the device.

The contributions of this paper are as follows:

- The proposed viscous torque assist method for the knee joint in the sitting posture was confirmed by measuring the surface electromyographic potentials.
- We compared the viscous command method (proposed in this paper) with time command and angle

command and compared and verified the assist effect.

Viscous assist overview

Assist by viscosity

As the angular velocity changes in accordance with the actual movement of a person, we considered the effect of the misalignment to be small. In this study, the viscous torque is expressed as follows when the joint torque due to viscosity is τ, the joint angular velocity is $\dot{\theta}$, and the viscosity coefficient is c.

$$\tau = c\dot{\theta} \tag{1}$$

Although it is difficult to extract the torque due to viscosity from the actual torque output by humans, the torque that varies with angular velocity, as shown in Eq. (1), is hypothetically considered to be the torque due to viscosity. In this study, the relationship between joint angular velocity and joint torque at the knee joint was obtained from motion analysis, and the slope of the graph was defined as the viscosity coefficient.

Next, we define the viscous-assisted motion. There are two types of motions: active and passive. Active movement is one in which the angular velocity and torque act in the same direction; thus, the torque acts in the direction of the joint movement. Passive movement is a motion in which the angular velocity and torque act in opposite directions; thus, the torque acts in the direction that stops a joint. Because the torque due to viscosity acts in the direction of stopping the joint, viscous assist uses passive movement. Figure 1 shows an image of the viscosity coefficient calculated from the relationship between the joint angular velocity and joint torque, as well as a diagram showing active and passive movements.

Existing research on assistive devices using viscosity includes an example in which viscosity was adjusted by an MR fluid and controllable impedance was applied to a hip joint [9], and an example in which viscosity was reproduced by an MR brake, and the effect on motion due to differences in the magnitude of viscosity was investigated [6]. Lower limb orthoses utilize viscous elements as dampers [10]. However, there are a few cases, such as the present study, that attempted to consider viscosity from the slope of the graph of the relationship between joint angular velocity and joint torque based on motion analysis data to widen the tolerance range against misalignment by viscosity and to improve coordination.

Characteristics of assist by viscosity command

This section describes the differences between assist methods based on viscosity commands and existing assist methods. Existing assistance methods based on

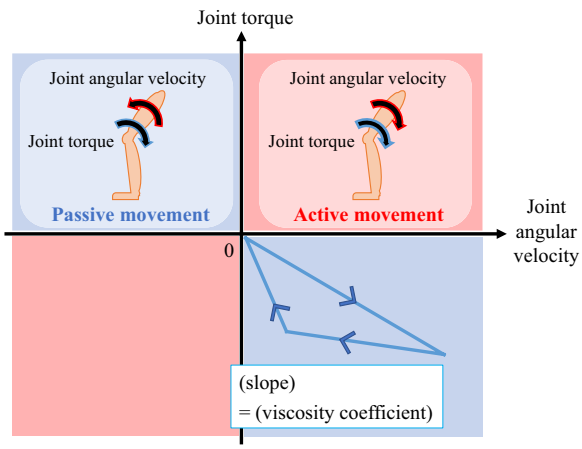


Fig. 1 Active movement and passive movement

feedforward control can be broadly classified into two types of motion generation:

Time command: A time command acquires command start information from a sensor or switch and outputs a predetermined command value in response to the passage of time. It is relatively easy to determine command values using time commands. However, it is difficult for the wearer to perform the operation in accordance with the set time every time, and there is a high possibility of a difference between the assumed and actual command timing.

Angle command: This command changes its value according to the equipment angle information. Unlike time commands, angle commands have a wide tolerance for deviation over time and are relatively easy to set because the command torque changes according to the angle of the device. However, there is a high possibility that correct assistance will not be provided in an environment that differs from the assumed environment (e.g., when the height of the chair differs in a seated movement) or when the motion trajectories of the device and wearer differ.

The frequency of command-switching varies; however, it is generally a combination of the above factors.

In contrast, viscous assist is a command whose command value changes according to the angular velocity information of the device and the command viscosity coefficient (hereinafter, referred to as "viscous command"). Because the viscous command changes its value according to the angular velocity information of the device, it is easier to match the actual motion of the wearer. We consider that focusing on angular velocity

allows a wider range of tolerance for deviation over time than time commands and a wider range of tolerance for environmental differences and deviation between the device and the wearer than angle commands.

In a previous study, we verified the command timing deviations for time, angle, and viscous commands [8]. The results showed that the viscous command had a lower command-timing deviation than the other methods. In contrast, this study evaluated viscous assist including the assist effect by surface EMG measurement using a prototype machine in which the assist force is increased by installing a planetary gear mechanism.

Motion analysis in seated movement

A previous study [8] showed that seating motion is an appropriate motion for viscous assistance using motion analysis. In this study, we determined the parameters for assistance based on motion analysis data. The same data as in the previous study were used for the motion analysis of the seating motion; however, the range of data obtained was different.

Experimental conditions in motion analysis

In the motion analysis, the subject sat on a 482 mm high platform. Reflective markers were attached to the subject's entire body, and their movements were measured using an optical motion capture system (MAC3D system, Motion Analysis, sampling frequency: 100 Hz). Two floor reaction force gauges (TF-4060-D, Tec Gihan) were placed on the subject's feet and the seated surface. The floor-reaction force gauges were synchronized with a motion capture system to measure the participants' floor-reaction force during movement. After measurement, the joint angles, angular velocities, and joint torques at the knee joint were obtained using musculoskeletal analysis software (n-Motion, NAC Image Technology, Inc.). The knee joint angle in the standing position was set at 0 deg, the direction of flexion was set to positive, and the direction of extension was set to negative. Data were acquired from the time the knee joint angle exceeded 1 deg until approximately 0.2 s after the joint angle reached 0 deg/s. Data were expressed as 0–100% from the start of movement to the end. The subject was a healthy male (age, 22 years, height 1.79 m, weight 60 kg), and data from five sessions were used.

Experimental results in motion analysis

Figure 2 shows the results of the motion analysis. The first row shows the results of the joint angle, the second row shows the results of the joint angular velocity, and the third row shows the results of joint torque. The vertical axis represents the joint angle, angular velocity, and torque, and the horizontal axis represents the movement

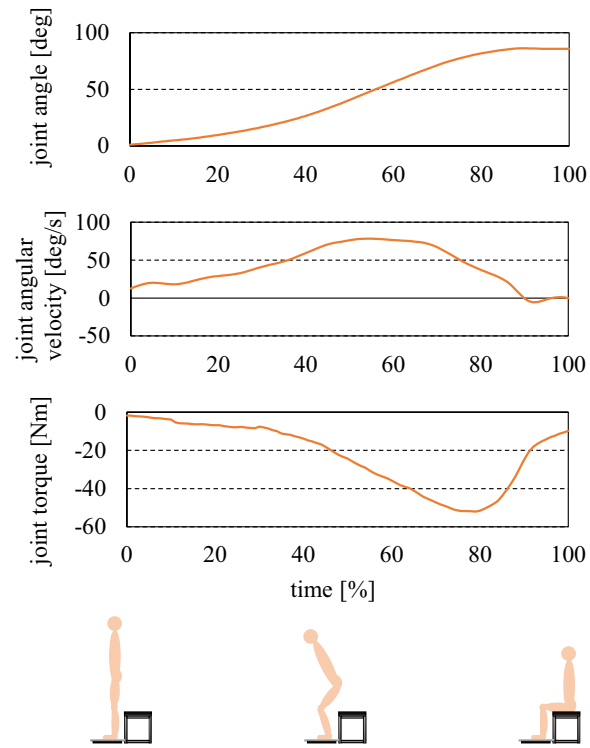


Fig. 2 Result of motion analysis

time. Figure 2 shows that the joint angle increased with time in the flexion direction. The joint angular velocity increased in the flexion direction up to approximately 55% and decreased thereafter. The absolute value of joint torque increased up to approximately 80% and decreased thereafter. Table 1 summarizes movement times. The mean operative time was approximately 2.0 s.

Figure 3 shows the relationship between the joint angular velocity and joint torque when the maximum output torque of the MR brake is considered. Because the maximum output torque of the MR brake is approximately 8 Nm, the MR brake can assist about 15.4% of the joint torque obtained from the motion analysis. Note, however, that in the experiments in this paper, the MR brake alone is not used, and the assist rate is increased by the planetary gear mechanism. Assist devices with planetary gears are described in Chapter 4. Therefore, the vertical axis in Fig. 3 represents 15.4% of the joint torque, and the horizontal axis represents the joint angular velocity. The slope of the approximate line in Fig. 3 is −0.0718. Because the direction of flexion is negative, the viscosity coefficient is 0.0718 Nm s/deg from Eq. (1). Figure 4 shows the time series of joint torque at 15.4% and viscous torque generated by multiplying the joint angular velocity by a

Table 1 Time of seated motion

Trial	1	2	3	4	5	Average
Time [s]	1.96	2.12	1.81	2.14	2.16	2.04

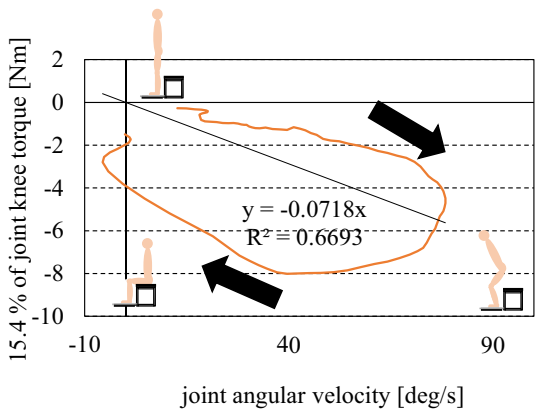


Fig. 3 Relationship of joint torque and angular velocity

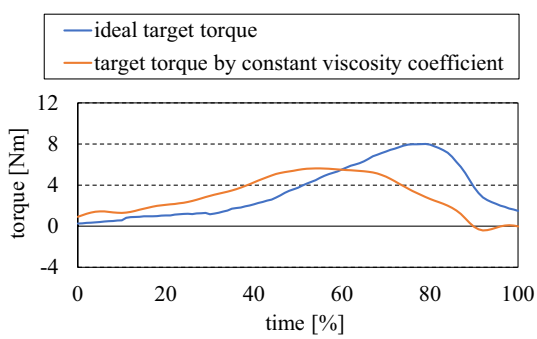


Fig. 4 Ideal target torque and target torque by constant viscosity coefficient

viscosity coefficient of 0.0718 Nm s/deg. Figure 4 shows that the actual peak values of the joint angular velocity and joint torque are different; however, they are similar in that they both have a peak shape, even though the viscosity coefficient remains constant throughout the entire motion. However, because the peak timing of torque and angular velocity are different, there is a difference between the peak timing of the target torque and the viscous torque. In order to bring the target torque and the viscous torque close, it is necessary to change the viscosity coefficient in accordance with the motion. However, since the viscous torque changes in accordance with the angular velocity during motion, we believe that even if there is a difference between the target torque and the viscous torque, the wearer would feel little discomfort.

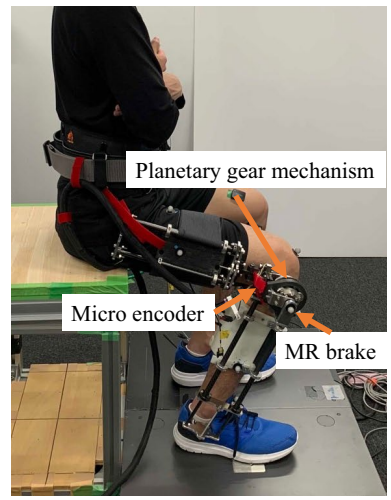


Fig. 5 Assistive device

Assist device with planetary gear system

Assist device

The knee-assist device used in this study is illustrated in Fig. 5. In this study, a viscous element was reproduced using a magnetorheological fluid brake (MR brake) (ER Tec). The torque owing to viscosity is expressed as the apparent viscosity by outputting the brake torque corresponding to the angular velocity. Details of the MR brake are described in the next section.

The assistive device consisted of an MR brake, a microencoder (Microtech Laboratory Co., Ltd., MES-6P Series Ver. 4), a planetary gear mechanism, attachments, thigh and lower leg arms, and a footplate. The weight of the device was approximately 2.5 kg. A microencoder was used to obtain device angle information. A planetary gear mechanism was attached to the output side of the MR brake to increase the output torque of the MR brake to the wearer. The worn part was fabricated using a 3D printer. Because the human thigh and lower leg have different thicknesses, different thicknesses were used for each part. The Velcro tape was used to fasten the wearer to the part. The worn parts were attached to the thigh and lower leg arms and transmitted the brake torque

from the MR brake to the wearer. The lengths of the thigh and lower leg arms can be adjusted according to the wearer. Ideally, the MR brake should approach the knee joint; however, in practice, misalignment occurs during the operation. In addition, if the weight of the device is secured to the wearer only with Velcro tape, the wearer's wearing area will be burdened, and a footplate will be installed to reduce the effect of the weight of the device. The device was attached to a belt. This structure avoids the weight of the device being applied only to the wearer's wearing part when the legs are raised and distributes the weight to the lower back as well.

Command signals to the device were output as voltages from the measurement and control system ControlDesk (dSPACE, Inc.) based on the information programmed in MATLAB/Simulink (2015, The MathWorks, Inc.). Subsequently, because the MR brake is current controlled, the voltage command is changed to a current command via the driver. The device-angle information acquired by the microencoder was recorded in the ControlDesk measurement and control system. The angular velocity information required for the viscosity command was calculated based on the angle information.

MR brake

In this study, MR brakes were used to reproduce the viscous elements. To compare the command methods in this study, we aimed to provide the wearer with as much commanded torque as possible. However, using a motor and reduction gear has the disadvantage that the inertia increases when sufficient assist torque is output. Therefore, an MR brake that can output high torque with low inertia was used. The MR brake used in this study has a maximum applied current of 0.5 A and a maximum output torque of approximately 8 Nm. A schematic of the MR brake is shown in Fig. 6. The interior of an MR brake contains a functional fluid called the MR fluid, whose properties change when a magnetic field is applied. MR fluid is mainly composed of three components: magnetic ultrafine particles, surfactant, and fluid as a medium. When a magnetic field is applied to an MR fluid, ultrafine magnetic particles form clusters in the direction of the magnetic field [11]. The multilayer disk then cuts the clusters as the core of the MR brake rotates relative to the case. The shear force generated during this process was the frictional torque of the MR brake. Because the response of the MR brake is as high as several tens of milliseconds, the frictional torque of the MR brake is controlled according to the angular velocity, thereby reproducing the torque due to viscosity.

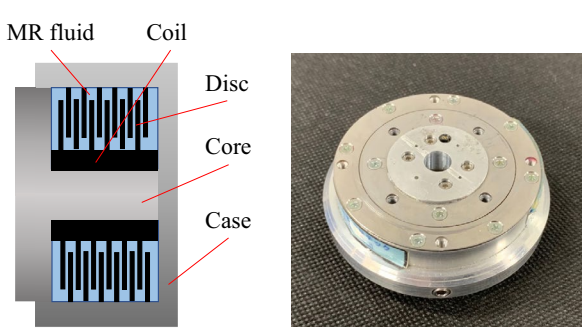


Fig. 6 MR brake

Output experiment of an assist device with planetary gear mechanism

In this section, we measured the increase in brake torque output to the wearer by installing a planetary gear mechanism. For this purpose, we compared the brake torque of an MR brake with that of an assist device equipped with a planetary gear mechanism. The brake torque of the MR brake alone was measured by rotating the MR brake with a motor and bringing an arm attached to the output side of the MR brake into contact with a fixed-load cell, as shown in Fig. 7. The rotation speed of the motor was set to 25, 50, 75, and 100 deg/s, and the command current

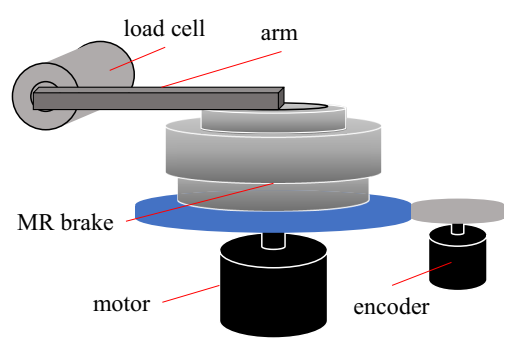


Fig. 7 Experimental device for unit of MR brake

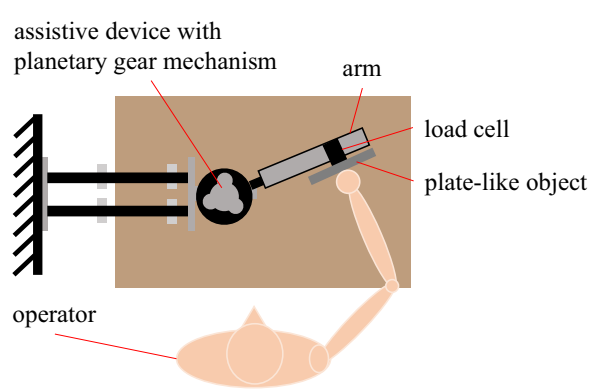


Fig. 8 Experimental device for assistive device with planetary gear mechanism

was measured from 0 to 0.5 A in 0.1 A increments. The braking torque of the planetary gear-mounted assist device was measured by pushing a load cell attached to the arm through the plate-like object, as shown in Fig. 8.

Figure 9 shows the output torque results at each command current for the assist device equipped with the MR brake and planetary gear mechanism. The vertical axis represents the output torque, and the horizontal axis represents the command current. The orange circles represent the results for the MR brake, and the blue squares represent the results for the planetary gearbox-equipped assist device. The output torque of the MR brake was assumed to be the average of the four rotational speeds at each command current because the difference in the rotational speeds was small. Figure 9 shows that the output torque of both the MR brake and the assist device with the planetary gear mechanism increased as the command current increased. The ratio of the increase in the output torque of the MR brake to that of the planetary gearbox assist device at each command current was 2.8 times on average. This value was lower than the assumed threefold increase in the original gear ratio. This is thought to be due to the energy consumption during torque transmission caused by friction and other factors. However, no significant differences were observed, and it was confirmed that the output torque increased more than that of the MR brake alone.

Measurement of muscle activity in seated movement

The purpose of this section is to measure the muscle activity at the knee joint during seated movements in three different ways of commanding the MR brake.

Control method

The three command methods—time command, angle command, and viscosity command—were set based on the results of the motion analysis described in Chapter 3. The motion time was set to 2 s, based on the motion analysis time obtained in "Motion analysis in seated movement" section. In addition, the work for each command was set to be approximately the same, based on the results of the motion analysis. Because the maximum output of the MR brake was approximately 8 Nm, the work was determined using a value of 15.4% of the joint torque; thus, the maximum value of the joint torque obtained from the motion analysis was approximately 8 Nm. The actual torque applied to the wearer was increased by the planetary gear mechanism. Although the increase owing to the planetary gear mechanism varies depending on the motion, command current, and other factors, this study assumes that the increase is approximately 2.8 times greater. Therefore, the increment due to the planetary gear mechanism is considered to be linear. Thus, there is no problem even if the increment due to the planetary gear mechanism is not considered when considering the work. The method used in this work is described below. Work W can be obtained from the angle θ and torque τ. Therefore, the joint angle in i % of the motion analysis is θ_i and the joint torque is τ_i, and the work W in this paper is expressed by the following equation:

$$W = \frac{1}{2}\sum_{i=1}^{100}((\theta_i - \theta_{i-1})(\tau_i + \tau_{i-1})) \tag{2}$$

The work of each command is obtained based on Eq. (2).

For the time command, the command value is a function of time because the command torque changes over time. The operation time was 2 s. The command start time was determined using information from a switch held in the subject's hand during measurement. In other words, the command begins when the switch is pressed. The work W_{time} obtained from the results of the motion analysis for the time command was $W_{time} \fallingdotseq 355$ J from Eq. (2). The set time command is shown in Fig. 10a. The vertical axis represents the commanded torque without considering the planetary gear mechanism, and the horizontal axis represents time.

The angle command is a function of the angle because the command torque varies with the equipment angle. In this case, the angle was adjusted to increase monotonically with device angle. Because the device angle at the completion of seating differed for each subject, the device angle at the completion of seating for each subject was obtained before measuring the angle command. The work W_{angle} in the angle command obtained from the motion analysis results is equal to the work W_{time} in the time command, so $W_{angle} = W_{time} \fallingdotseq 355$ J.

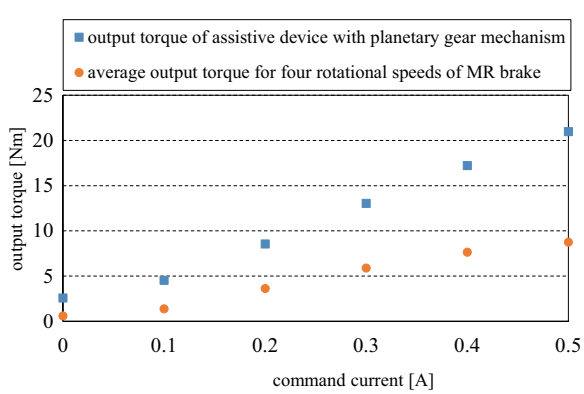


Fig. 9 Result of output torque at each commanded current for MR brake and assistive device with planetary gear mechanism

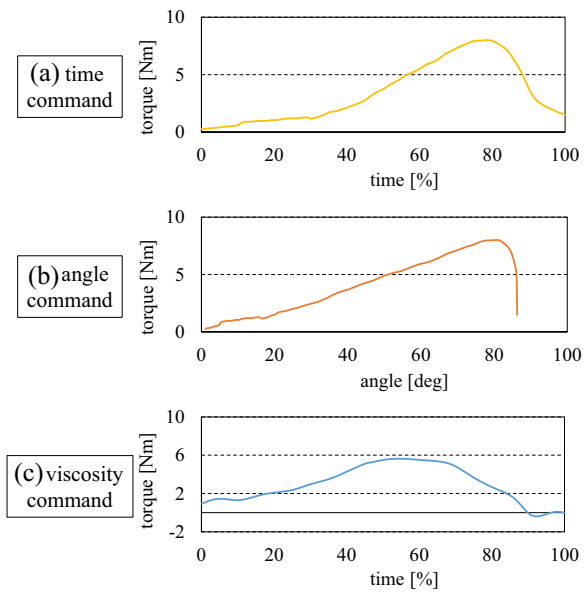


Fig. 10 Command torque of time command, angle command and viscosity command

The set-angle command is shown in Fig. 10b. The vertical axis represents the command torque, which does not consider the planetary gear mechanism, while the horizontal axis represents the angle.

The viscosity command is a function of the angular velocity because the command torque depends on the angular velocity of the device and the command viscosity coefficient. However, the viscosity coefficient does not change in this experiment. The command viscosity coefficient, c, was set such that the work W_{time} in the time command, W_{angle} in the angle command, and $W_{viscosity}$ in the viscosity command were approximately the same. From Eqs. (1) and (2), $W_{viscosity}$ can be expressed as

$$W_{viscosity} = \frac{1}{2}\sum_{i=1}^{100}\left((\theta_i - \theta_{i-1})\left(c\dot{\theta}_i + c\dot{\theta}_{i-1}\right)\right). \quad (3)$$

Because the joint angular velocity $\dot{\theta}$ at each movement time is determined by motion analysis, it can be expressed as

$$W_{viscosity'} = \frac{1}{2}\sum_{i=1}^{100}\left((\theta_i - \theta_{i-1})\left(\dot{\theta}_i + \dot{\theta}_{i-1}\right)\right) \quad (4)$$

to obtain

$$c = \frac{W_{time}}{W_{viscosity'}}, \quad (5)$$

The command viscosity coefficient c can be obtained. Therefore, from Eq. (5), the command viscosity is 0.0718≒0.072 Nm s/deg. Considering the planetary gear mechanism, the viscosity is 0.20 Nm s/deg. At this time, $W_{viscosity} \fallingdotseq 352$ J, which confirms that the work is approximately the same for each command. Furthermore, c = 0.072 Nm s/deg, which is approximately equal to the slope of the approximate straight line obtained for the relationship between joint torque and joint torque of 15.4%. The set viscosity commands are shown in Fig. 10c. The vertical axis represents the command torque without considering the planetary gear mechanism, and the horizontal axis represents time.

Experimental environment

The subjects sat on a chair at a height of 482 mm. Floor reaction force gauges were placed on the subject's right and left feet, and the subject held a switch in his hand. The subjects pressed the switch when starting the movement, and the movement started as soon as the switch was pressed. The assist device was attached to the subject's right leg, and the muscles measured were the left and right vastus medialis and left and right semimembranosus. The vastus medialis generates torque in the direction of knee extension, and the semimembranosus generates torque in the direction of knee flexion. As shown in the motion analysis results, the knee joint in the seated motion was flexed while supporting the body weight; therefore, the torque acted in the direction of extension. Therefore, we assumed that the vastus medialis was primarily active. Alternatively, we assume that the muscles in the direction of extension and those in the direction of flexion are active against each other, stiffening the knee motion. An EMG sensor (Delsys Trigno, Delsys) was used to measure the EMG. The subjects were five healthy males (mean age: 22.6 ± 1.3 years, mean height: 176.4 ± 4.9 cm, mean weight: 63.2 ± 9.2 kg).

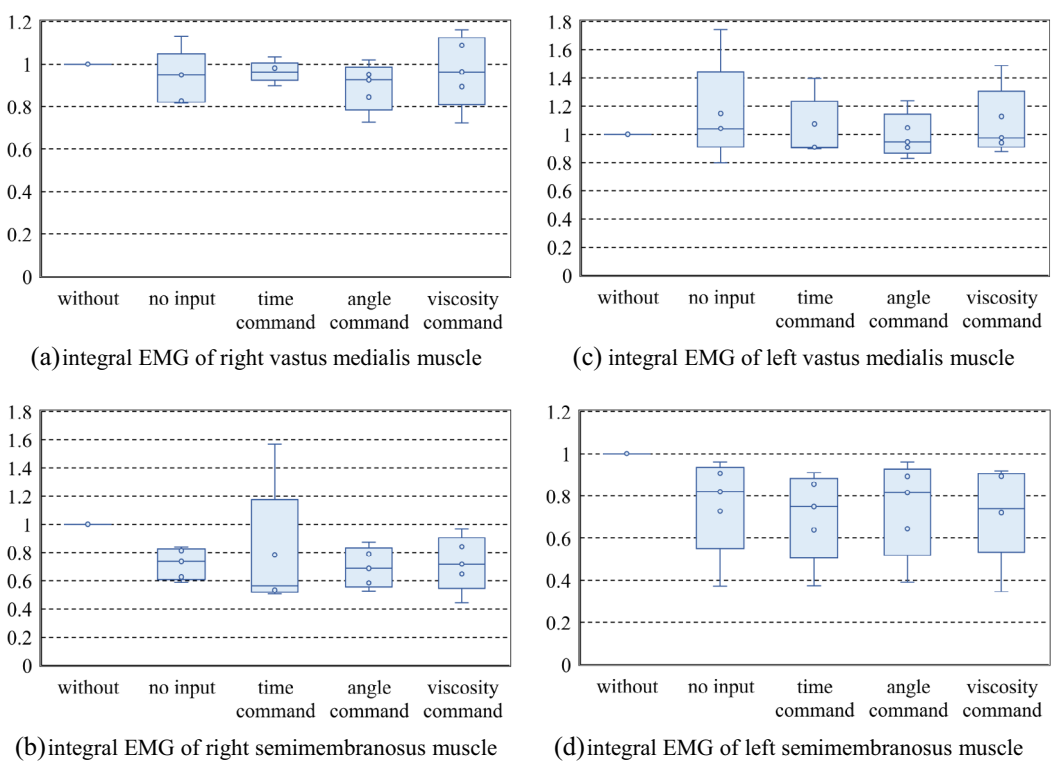
Experimental procedures

There are a total of five measurement conditions. The order of measurement was as follows: without the assistive device (without), with the assistive device installed but with no command applied (no input), with a time command applied, with an angle command applied, and with a viscosity command applied. Before measuring the angle command, the angle at the completion of seating was measured for each participant. The motion duration was set to 2 s based on the motion analysis results, and a metronome was used to indicate the motion duration. The MR brake used in this study is similar in structure to the MR brake used in Okui [12]. The time constant of the MR brake is 0.025 and the rise time is 54 ms [12]. The sampling rate of the encoder is 2 kHz. The data were processed after the experiment. The data used for data

processing were obtained approximately 0.2 s after the value of the floor reaction force gauge placed at the right leg fell below a certain value from the timing at which the subject pressed the switch in his hand. The measured EMG data were bandpass filtered (20–450 Hz) and full-wave rectified. The data were then low-pass filtered (10 Hz) and normalized to a time series of 0–100%. The integral myopotential values were also calculated by integrating the acquired myopotential data from 0 to 100% of the movement interval. In this experiment, data from eight trials were used for each condition, and the average value was used. The integrated EMG for each measurement condition was then divided by the integrated EMG for the "without" condition to make the "without" condition the reference. Data processing included the processing of myopotential data, ControlDesk input/output data, and data from switches and floor reaction force plates, with a maximum time difference of approximately 11 ms. However, this was negligible compared to the movement time and was therefore ignored. The Ethical Review Committee of Chuo University approved this study (No. 2021-064).

Results of experiments

Figure 11 shows a box-and-whisker plot of the integrated myopotentials of the left and right vastus medialis and semimembranosus muscles for the five subjects. The quartiles were calculated using the exclusive median. The small circles in Fig. 11 show the second, third, and fourth EMG values for each trial from the top. Note that for each measurement condition, each subject was divided by the value of the without integral EMG for each subject to use as a reference. In addition, the meaning of the median value in this paper indicates the middle value. The result in Fig. 11a for the vastus medialis of the right leg shows that the median value was lower than the median value for all command methods. Because there were five subjects in this experiment, the fact that the median value was lower than the without indicates that more than half of the subjects had lower integrated EMG potentials than those without. However, the maximum value for each command condition was higher than that without commands. Based on these facts, it is thought that more than half of the subjects experienced an assist effect in the vastus medialis muscle in any command method; however, some subjects did not experience an assist effect. The assist effect was not obtained for the viscosity command of Subject C, angle command of Subject D, and time and viscosity commands of Subject E. The maximum assist effect was greater than that without the command. In addition, the largest decrease in the EMG was observed



(a) integral EMG of right vastus medialis muscle

(c) integral EMG of left vastus medialis muscle

(b) integral EMG of right semimembranosus muscle

(d) integral EMG of left semimembranosus muscle

Fig. 11 Box-and-whisker plots of integral EMG based on data of without

in the viscosity command of Subject B when compared with the without.

Figure 11b shows the results for the right semimembranosus muscle. The median value was smaller than that without all command conditions. The median value was smaller than the median value of without for all conditions, except for the time command. However, only the time command of subject E showed an increase in the EMG potentials compared to that without the command. This is thought to be due to the misalignment of the assist timing, which causes a demand for force in the knee flexion direction. Therefore, we consider that the time command may cause an increase in myopotential owing to a deviation in command timing; however, other command methods can assist the semimembranosus muscle. In addition, since the potential of the semimembranosus muscle decreased, we believe that the antagonist muscle was also active and that the assist suppressed the activity of the antagonist muscle.

Figure 11c shows that the median value of the vastus medialis of the left leg did not change significantly from that without the device; thus, we consider that the effect on the left leg without the device is small. However, the median value of no input was larger than that of without, and the maximum value was also the largest. This is because the no input was not commanded to the device and the force received from the device was small, so we believe that the weight could not be successfully placed on the device and the EMG potentials increased. The time command, angle command, and viscosity command of subjects D and E increased the EMG potential more than without. We believe that this is because the subject was not accustomed to being assisted, and thus generated unnecessary force.

Figure 11d shows that the EMG signal of the semimembranosus muscle of the left leg was lower for all command methods. This suggests that the device did not adversely affect the semimembranosus muscle of the left leg that was not wearing the device.

Next, the assumed and actual command signals for each command method are discussed. Figure 12 shows the change in the command torque for each subject under each measurement condition. The changes in the command torque for subjects A, B, C, D, and E are shown from the top. The vertical axis represents the command torque, and the horizontal axis represents the operation time. The red dotted line shows the assumed command values for the time axis in the time and angle commands; the blue dotted line shows the assumed command values for the time axis in the viscosity command; the yellow solid line shows the actual command values in the time command; the orange solid line shows the actual command values in the angle command; and the blue

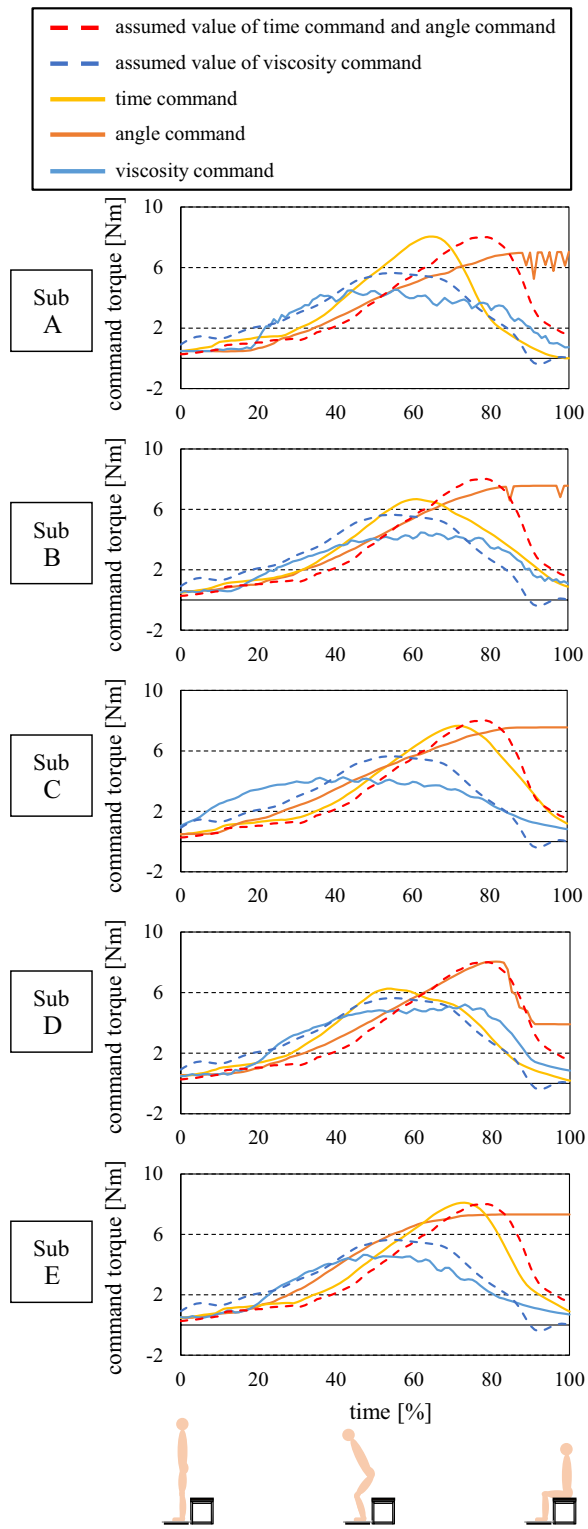


Fig. 12 Assumed command signal and actual command signal

solid line shows the actual command values in the viscosity command. Figure 12 shows the average of eight measurements. First, for the time command, some subjects showed a transition close to the assumed command value, whereas others showed a difference between the assumed and actual command values. In this experiment, a metronome was used, and the timing was adjusted so that the subjects could easily adjust the timing. However, as shown in Fig. 12, there was a discrepancy. This suggests that we consider that it is difficult for the time command to perform actual movement in accordance with the assumed command value. Next, in the angle command, the command voltage does not decrease beyond 80%, in contrast to the assumed command value. This is considered to be because the correct command voltage was not applied owing to the misalignment between the device and the wearer. In this experiment, the angle at the completion of seating was measured and adjusted for each subject, but the effect of misalignment between the device and wearer was apparent. In addition, the command value oscillated after 80% for Subjects A and B. This is because the device is set not to output a command value when the angular velocity is in the direction of extension, and the average value of eight times is acquired. We believe that the reason why the angular velocity is in the direction of extension is due to the misalignment between the device and the wearer. However, since the majority of the seating motion is completed during the oscillating portion, the effect of this on the amount of muscle activity is considered to be small. This suggests that we consider that it is difficult for the wearer and the device to operate without misalignment and that it is difficult to apply the assumed command value. Next, in the viscosity command, there was no significant discrepancy between the assumed and actual commands, except for subject C. We believe that the reason for the discrepancy between the assumed command signal and the actual command signal in subject C is that he started moving before pressing the switch. However, the other four subjects did not show large discrepancies between the assumed and actual command signals. This is because, unlike the time command, the angular velocity of the device changed accordingly, even when the motion time deviated; therefore, we consider that no large deviations occurred. Furthermore, unlike the angle command, the command voltage did not reach a high level after 80%. This is because we focused on the angular velocity; thus, we believe that the device was not significantly affected by the misalignment between the device and wearer. Based on the above, we believe that the viscosity command is less susceptible to the effects of time lapse and misalignment between the device and the wearer than the time and angle commands.

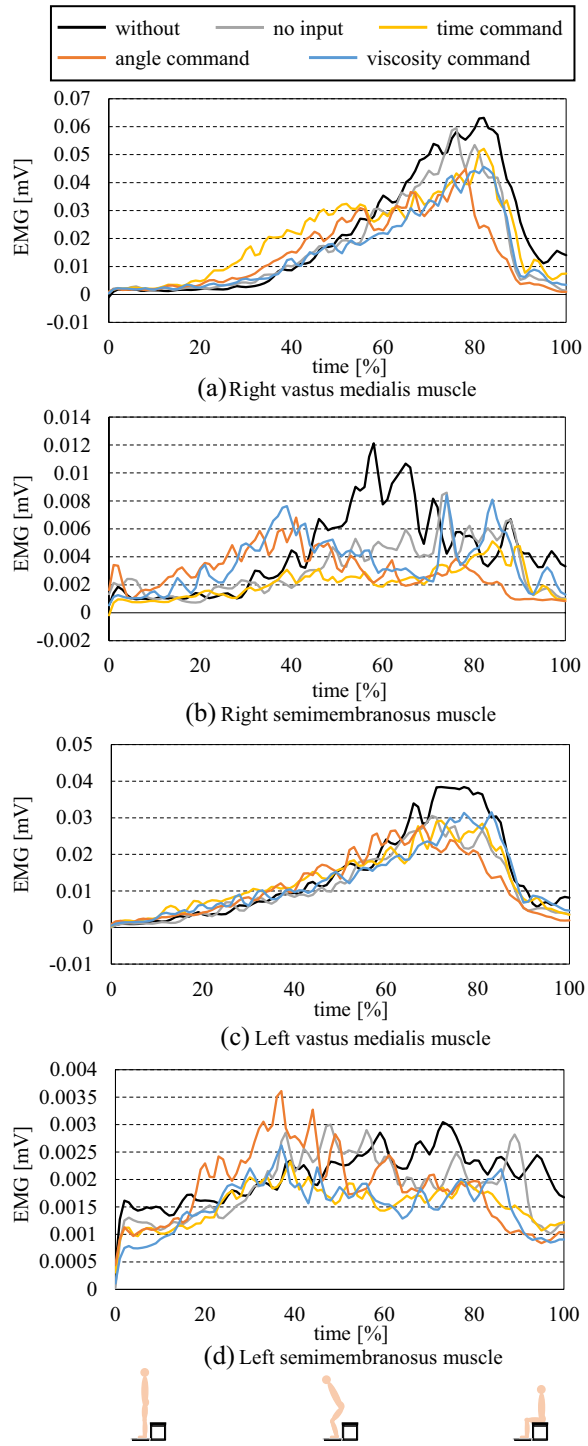


Fig. 13 EMG of Subject C

The time-series EMG potentials of Subject B, whose EMG potentials decreased the most owing to the assist effect, and Subject E, whose EMG potentials increased in the vastus medialis and semimembranosus muscles, are discussed further. Figure 13 shows the results of the

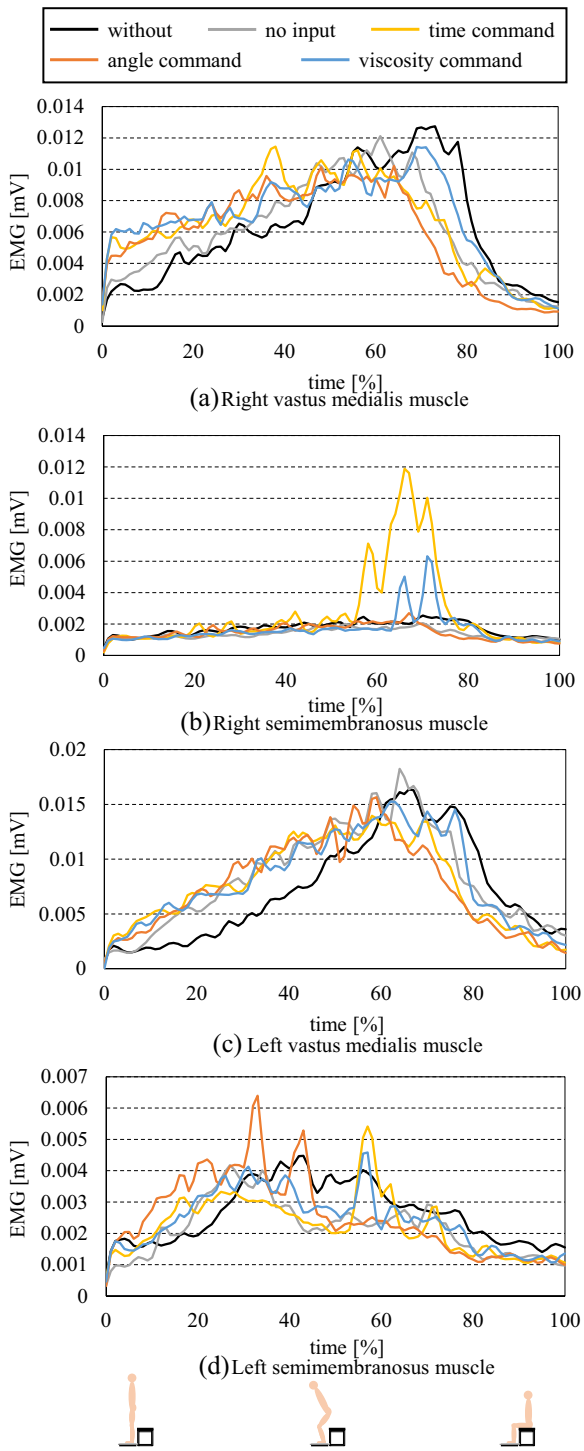


Fig. 14 EMG of Subject E

measurement of the EMG potentials of subject B, and Fig. 14 shows the results of the EMG potential measurements of subject E. In both cases, the vertical axis represents the muscle potential, and the horizontal axis represents the movement time; the data are for the right vastus medialis, right semimembranosus, left vastus medialis, and left semimembranosus from the top. The black, gray, yellow, orange, and blue solid lines represent the without, no input, time, angle, and viscosity commands, respectively. First, we discuss the results for Subject B in Fig. 13. The medial vastus medialis of the right leg had a peak value around 80% under all measurement conditions. Compared with the without, there was a decrease in EMG potential around 60–100% for all command methods except no input. EMG potentials of no input were higher than those of without at around 70%, but were lower than those of without at the rest of the range. According to Fig. 9, a constant torque was output even when no command was applied to the MR brake, so we believe that there was an assist effect. However, when no command was applied to the MR brake, the output torque of the device was small, and the decrease in EMG potential was smaller than that of the time command, angle command, and viscosity command at around 60–80%. In addition, the time and angle commands increased the EMG potentials in the 20–55% range compared to the without command, but there was no increase in the EMG potentials in the viscosity command. Therefore, we consider that the assist effect of the viscosity command was obtained in the medial vastus medialis of Subject B. In contrast, in the semimembranosus muscle of the right leg, the without command case led to an increase in the myopotential around 50–70%. The angle command and viscosity command EMG potentials increased at approximately 20–40%. This is thought to be due to an increase in the EMG potential of the semimembranosus muscle that flexes the knee as the knee flexor attempted to sit up within the time by accelerating the knee because the movement time was set during the assist. However, the integrated myopotential was reduced during the entire movement compared with without, suggesting that the entire movement was assisted. In addition, although this experiment was conducted at a constant coefficient of viscosity, we believe that a further assist effect can be obtained by a variable coefficient of viscosity. The results for the vastus medialis and semimembranosus muscles of the left leg showed no increase in EMG potentials compared with those without and no significant differences in each command, suggesting that the effect on the left leg was small. Next, the results for Subject E in Fig. 14 are described. The medial vastus medialis of both the right and left legs showed an increase in the EMG potential of each command in the 0–60% range compared with the without. However, the EMG potentials in both legs decreased from 60 to 100%. In particular, the maximum value of the EMG potential for each command of the right leg was lower than the maximum value without a command. The increase in EMG potentials in the first

half of the movement is considered to be due to the fact that the subject was not accustomed to the assist device, as EMG potentials increased with the start of the movement, and thus the subject was unable to be assisted properly and exerted unnecessary force. On the other hand, there was no significant difference in the semimembranosus muscle of the left leg; thus, we considered that the effect of the assist was small. In the right semimembranosus muscle, time command EMG potentials increased by around 55–75%, which roughly corresponds to the area where the command voltage is higher than that in Sub E of Fig. 12. Therefore, we consider that the subject generated a force in the direction to overcome the assist force and that it was difficult for subject E to be assisted because the time command tended to deviate from the command value. In this experiment, a metronome was used to make the target movement time easier to understand. We believe that this caused the subject to forcibly apply force against the assist force to match the metronome, which generated force in the direction of flexion. We believe that this was caused by the fact that Subject E was not accustomed to the sensation of being assisted. From the above, we believe that it was difficult for subject E to be assisted because the time command tends to deviate from the command value.

Conclusion

In this study, we propose an assistive method using the viscosity calculated by the viscosity coefficient and angular velocity to improve the coordination with human movement. We measured the amount of muscle activity using surface electromyography in an assistive experiment using a lower-limb exoskeleton-type assistive device with an MR brake. To verify the assisting effect of the viscous command, a comparison was made between the time and angle commands. More than half of the subjects showed a decrease in myopotential in any of the command methods compared with the without. The largest decrease in the myopotential was observed for the viscosity command of Subject B. An actual decrease in myopotential was also observed in the viscosity command of Subject B. The results showed that the viscosity command of Subject B was the most effective for all subjects. By comparing the assumed command signal with the actual command signal, we found that the viscosity command was closer to the assumed command signal than the time or angle commands. These results confirm that viscous assist is less affected by deviations in time and device angle and has an assist effect. In this experiment, the viscosity coefficient was constant throughout the entire movement, resulting in a difference between the target torque and the viscous torque. However, we believe that in the future, by making the viscosity coefficient variable, the difference between the target torque and the actual torque can be reduced and a more coordinated assist can be provided. In the future, we plan to provide assistance for complex movements other than seated movements such as descending stairs.

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Author contributions
YS conducted the viscosity assist experiments. TS, MO, RN, and TN discussed the study and supervised the writing of the manuscript. All authors read and approved the final manuscript.

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Declarations

Competing interests
The authors declare that they have no competing interests.

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