Development of an orthosis for walking assistance using pneumatic artificial muscle

A quantitative assessment of the effect of assistance

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Abstract—In recent years, there is an increase in the number of people that require support during walking as a result of a decrease in the leg muscle strength accompanying aging. An important index for evaluating walking ability is step length. A key cause for a decrease in step length is the loss of muscle strength in the legs. Many researchers have designed and developed orthoses for walking assistance. In this study, we advanced the design of an orthosis for walking assistance that assists the forward swing of the leg to increase step length. We employed a pneumatic artificial muscle as the actuator so that flexible assistance with low rigidity can be achieved. To evaluate the performance of the system, we measured the effect of assistance quantitatively. In this study, we constructed a prototype of the orthosis and measure EMG and step length on fitting it to a healthy subject so as to determine the effect of assistance, noting the increase in the obtained step length. Although there was an increase in EMG stemming from the need to maintain body balance during the stance phase, we observed that the EMG of the sartorius muscle, which helps swing the leg forward, decreased, and the strength of the semitendinosus muscle, which restrains the leg against over-assistance, did not increase but decreased. Our experiments showed that the assistance force provided by the developed orthosis is not adequate for the intended task, and the development of a mechanism that provides appropriate assistance is required in the future.

Keywords—walking assistance; straight-fiber artificial muscle

I. INTRODUCTION

In recent years, the increase in the number of elderly people and those requiring care has become a social problem owing to the decline in the walking ability with age [1], [2], which decreases markedly during one's 60s and 70s [3]. This decrease diminishes opportunities for maintaining physical activity, which further advances the decline in the walking ability. An important index for evaluating walking ability is step length. Research has shown that step length decreases for various reasons, the loss of leg muscle strength being an important reason [4], [5]. To address this problem, in this study, we develop an orthosis to support leg muscle strength and to increase step length.

Many researchers have designed and developed orthoses for the same purpose [6], [7], [8]. For example, Honda developed an assist suit with motors that are controlled by information obtained from sensors located at the hip joints [6]. However, the motors have a high degree of stiffness and can pose a danger if not controlled properly. For this reason, we used a pneumatic artificial muscle as an actuator for an assist orthosis. This actuator has a low degree of stiffness.

We succeeded in constructing an orthosis for walking assistance on the basis of application of the pneumatic artificial muscle and evaluated the performance of the orthosis by measuring the effect of assistance quantitatively.

II. SYSTEM CONCEPT OF THE ORTHOSIS

The orthosis was designed for people with a slight gait disorder, such as elderly people, to improve their walking abilities by providing assistance and to further recuperate their walking abilities. To increase step length, assistive power is provided in the direction that flexes the hip joint at the time of the forward swing of the leg. Assistance is provided during all segments of the swing phase except for the downward swing of the leg. An image of the assistance timing is shown in Fig. 1. Assistance is added not to the hip extension but to only the hip flexion, in order to perform relatively efficient assistance with low energy and simple mechanism.

Owing to the force supplied by the artificial muscle, flexible assistance with a low rigidity can be provided, and the configuration can be wearable without the need for a hard external skeleton. As a result, the support system can be used comfortably in everyday life and assistive power suitable for human body is provided as required.

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A small compressor and a battery required for driving the orthosis is contained in the unit so that it can be powered independently for a long period. This encourages further increases in the sphere of a user's activity.

I encourage the people to use the orthosis in everyday life including airing as training and lead active life.

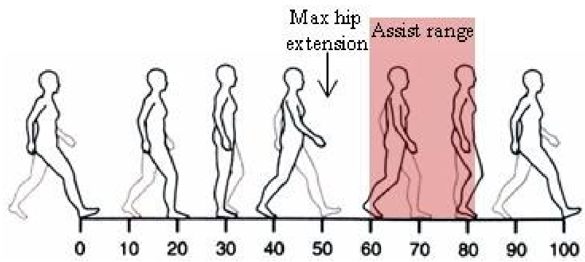

Fig. 1. Image showing assistance timing.

III. Composition of the Orthosis

Fig. 2 shows the appearance of the developed orthosis. The unit provides assistance for swinging the legs forward and upward during walking by the contraction of artificial muscles that run from the abdominal region to the knee region.

To provide assistance at an appropriate timing, the orthosis senses the angle of the hip joints by using potentiometers.

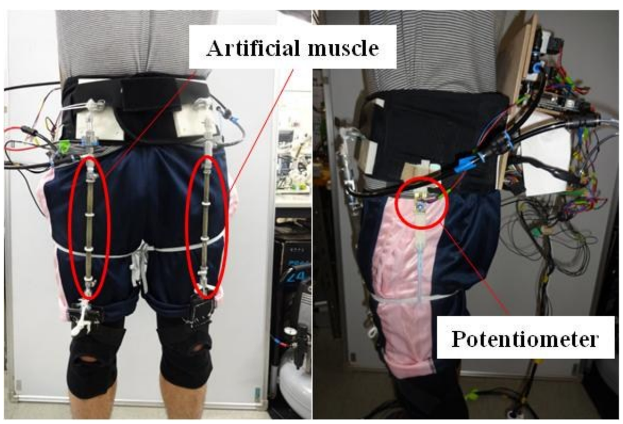

Fig. 2. Appearance of the orthosis.

A. Straight-Fiber Pneumatic Artificial Muscle

Fig. 3 shows a schematic diagram of a straight-fiber artificial muscle. The tube shown in this figure is made from natural latex rubber with a carbon-fiber sheet, which is inserted in the long-axis direction, and is secured at either end by terminals. Because this structure suppresses expansion in the axial direction, the artificial muscle, driven by air pressure, expands radially and exerts a contractile force axially.

The straight-fiber muscle is better than the McKibben muscle in terms of contractile force and the rate of contraction [10].

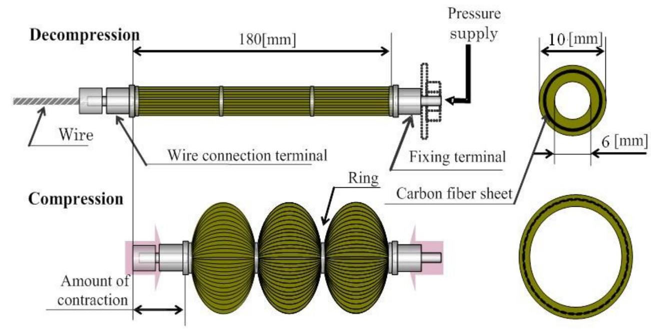

Fig. 3. Straight-fiber pneumatic artificial muscle.

In the design of an artificial muscle, the key parameters are contractile force and the amount of contraction. The parameters used in this study were determined as described below.

The inner torque of the hip joint in the sagittal plane becomes maximum in the early stages of the swing phase and is approximately 1.3 Nm/kg [9]. As the average weight of an elderly person is approximately 60 kg, inner torque is calculated at approximately 78 Nm. Because too great power will generate stretch reflex in the antagonistic muscle, the contractile force is set to 40 N so that nearly 5% of the inner torque can be assisted tentatively from this value.

The amount of contraction is established with reference to the simulation result shown below. As shown in Fig. 4, the motion of the orthosis and the subject is modeled in the sagittal plane.

The relationship between the angle of the hip joint in a walking cycle and the amount of slack from the maximum length of the artificial muscle is shown in Fig. 5. The general dimensions of the body type are used for calculation [9]. The graph shows that the amount of slack is at least 10 mm. Here, it is set to 40 mm, which is considered to be an adequate amount of contraction of the artificial muscle.

As mentioned above, the artificial muscle is designed such that it contracts 40 mm with a load of 40 N.

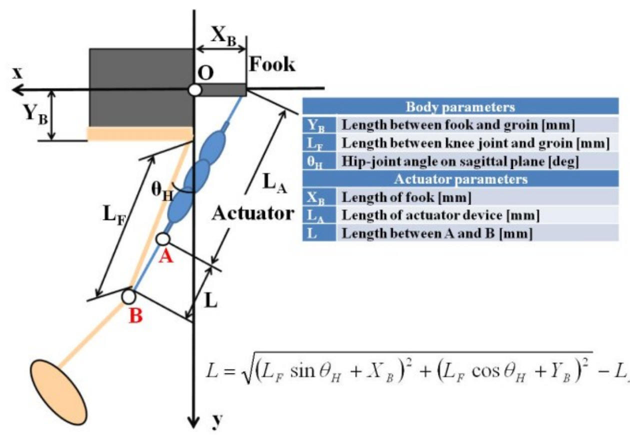

Fig. 4. Model of body in the sagittal plane.

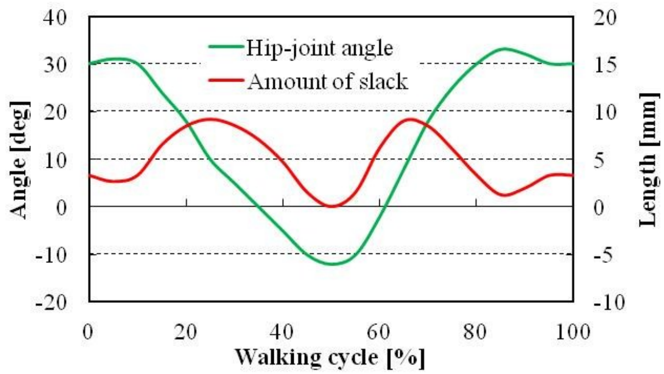

Fig. 5. Amount of slack in a walking cycle.

B. *Control Method of Pressure*

In the developed orthosis, as shown in Fig. 7, pressurized air from a tank passes through a pressure control unit and is supplied to the artificial muscle.

In the pressure control unit, there are two ON/OFF valves, which open or close the flow channel. These valves are used for applying and discharging air, respectively. The orthosis controls the air pressure by controlling these valves using a PWM signal.

A PWM signal is an electrical signal that repeats an ON/OFF cycle, as shown in Fig. 6. Its modulation is defined by the frequency and duty cycle of the signal. The PWM signal is characterized by frequency (Hz) and duty cycle (%).

In the developed orthosis, the pressure is controlled by a dual pneumatic control system (DPCS) that we developed.

1) DPCS System: The PWM signal (Fig. 6) is given to the DPCS system, as shown in Fig. 7. The parameters used for the derivation of the theoretical formula are shown in Table I.

Assuming "Q" is always a chalk flow proportional to the constant "a," the pressure of the artificial muscle can be expressed as shown in (5).

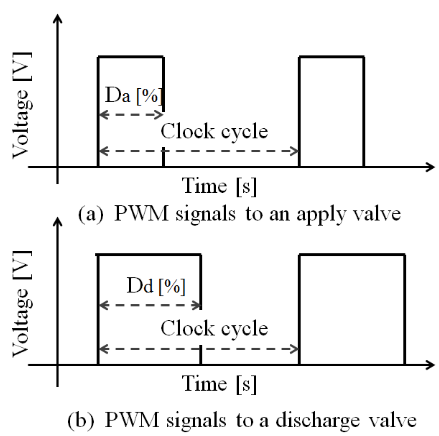

Fig. 6. Schematic of PWM signals for DPCS.

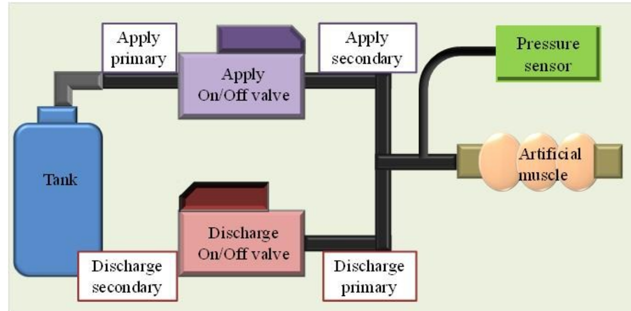

Fig. 7. Schematic of pneumatic apparatus.

TABLE I. PARAMETERS

Parameter	*Apply valve*	*Discharge valve*
Air content of a cycle [l]	A_a	A_d
Duty cycle [%]	D_a	D_d
Volumetric Flow [l/s]	Q_a	Q_d
Primary pressure [MPa]	P_a	P_d
Frequency [Hz]	f	

$$A_a = Q_a \times D_a \times 1/f \tag{1}$$

$$A_d = Q_d \times D_d \times 1/f \tag{2}$$

$$Q(P) = a \times P \tag{3}$$

$$a \times P_a \times D_a = a \times P_d \times D_d \tag{4}$$

$$P_d = P_a \times D_a/D_d \tag{5}$$

As mentioned above, the pressure of the artificial muscle is determined by the pressure in the tank and the proportion of the duty cycle for the applying and discharging valves. The use of DPCS enables a stable pressure control that is suitable for the artificial muscle and can decrease air consumption.

2) Respondence of constriction: Fig. 8 shows the experimental result of contracting the artificial muscle by means of DPCS pressure control according to an actual walking cycle. An experimental load of 40 N is applied to the artificial muscle, and the pressure in the tank is 0.25 MPa.

Pressure control is performed as follows:

- Air at tank pressure is supplied to an artificial muscle directly.

- Just before the pressure in the artificial muscle reaches target pressure, start DPCS to control pressure.
- When it reaches the target pressure, the pressure of the artificial muscle is maintained by closing each ON/OFF valves.
- Air is discharged after assistance has been provided. The pressure of the artificial muscle is maintained when it falls below the initial pressure. The initial pressure is set as maximum so that the artificial muscle does not contract for the purpose of increasing the artificial muscle's response speed.

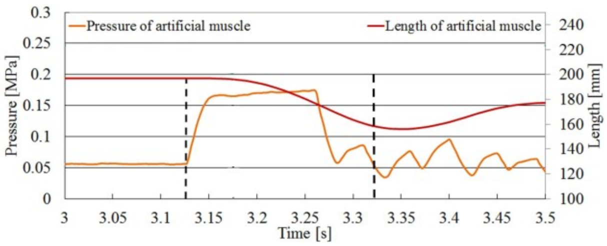

Fig. 8. Respondence of constriction.

The graph above shows that although pressure control can be performed, there is a time lag of approximately 0.20 s between the start of pressurization and the constriction of the artificial muscle. It is thought that the time lag has a significant influence on assistance provided compared with a general walking cycle of period 1.0 s. Therefore, pressurization is started ahead of the time lag, so that the artificial muscle can contract at the desired time.

C. Prediction of Walking Cycle

The average of the last three walking cycles, which depends on the data supplied from potentiometers, is calculated as the next walking cycle, and the delivery of pressurized air can be started ahead of the desired assistance timing.

The software program detects the local minimum of signal waveform from the potentiometer when the hip joint is completely extended. The walking cycle is the time between two local minima. The direction to start pressurization is impressed so that assistance is executed at a certain percentage of the walking cycle after the maximum extension of the hip joint. Here we set this value to 5%.

IV. Evaluating the Assistive Effect

We constructed a prototype of the orthosis for walking assistance and tested its usefulness by fitting it on a healthy subject and measuring its assistive effect quantitatively. Step length and EMG were measured as parameters for evaluations in this experiment.

A. Experimental Method

The subject was a healthy 22-year-old male walking on a treadmill at a rate of 3.5 km/h, which is the general walking speed. Experiments were conducted for assisted walking and unassisted walking without wearing orthosis that is normal walking. Assistance was provided to the right leg only owing to the experimental environment. The artificial muscle was fixed in the strained condition when the hip joint is completely extended so that the assistive force was provided efficiently.

Step length was estimated from coordinate data of the ankle joint on the sagittal plane by dynamic image analysis.

An EMG evaluation of the sartorius muscle and the semitendinosus muscle was conducted. The sartorius muscle, which provides flexion of the hip joint and the knee joint, swings the leg forward. The semitendinosus muscle, which provides extension of the hip joint and flexion of the knee joint, swings the leg backward prior to the heel contact. From the EMG data on these muscles, we analyzed the effect of assistance and the existence (or nonexistence) of action to restrain the leg rearward from over-assistance. The EMG signal was initially subject to bandpass filtering (30–300 Hz), root mean square (RMS) processing, and lowpass filtering (10 Hz), after which it was divided by the maximum voluntary contraction (MVC) for normalization.

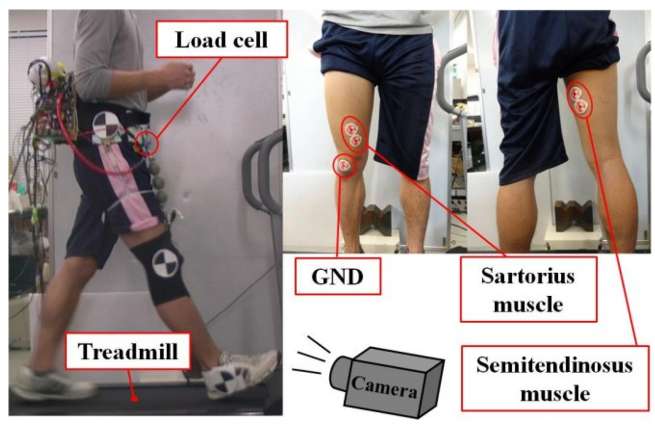

Fig. 9. Experimental setup.

B. Experimental Result

The average step length from animation analysis is shown in Table II. The transition of the coordinate on the x, y-axis on the sagittal plane of an ankle joint is shown in Fig. 10 (the x-axis is the direction of walking, while the y-axis is that of vertical rise). These data are the averages of ten walking cycles.

In this prototype, as the program needs to learn how to identify potentiometer noise, it is difficult to provide assistance at the appropriate timing each time. Therefore, we take the average value of ten walking cycles in which appropriate timing assistance is conducted. Fig. 11 is a graph of the pressure in the artificial muscle and tensile force measured by a load cell. Fig. 12

shows a graph of the tensile force and the angle of the hip joint. Figs. 13 and 14 show the EMG data with tensile force. The hip-joint angle, which is detected by the potentiometer, is 0° when the subject is standing, with flexion set to positive and extension set to negative.

TABLE II. COMPARISON OF STEP LENGTH

Normal Walk	Assist with orthosis
901.4 [mm]	924.6 [mm]

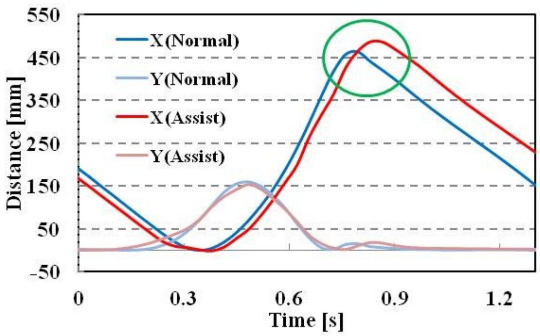

Fig. 10. Transition of ankle joint.

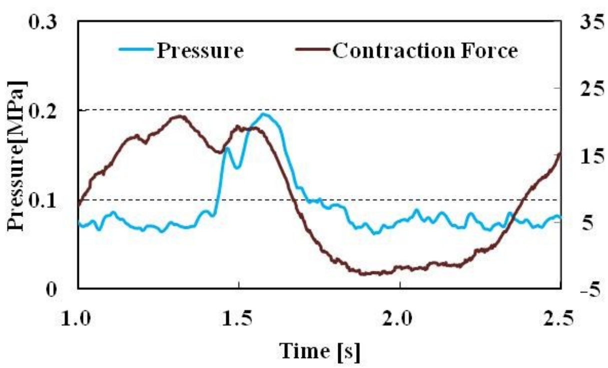

Fig. 11. Pressure in muscle and contractile force.

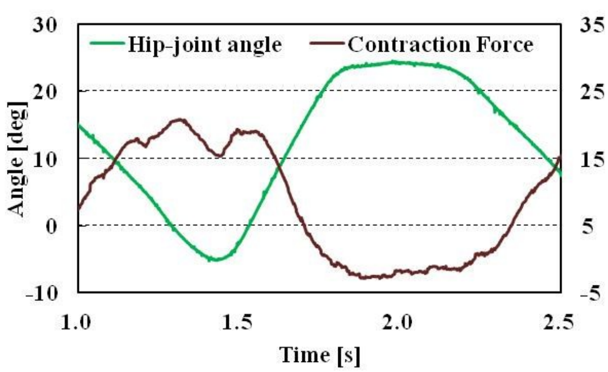

Fig. 12. Hip-joint angle and contractile force.

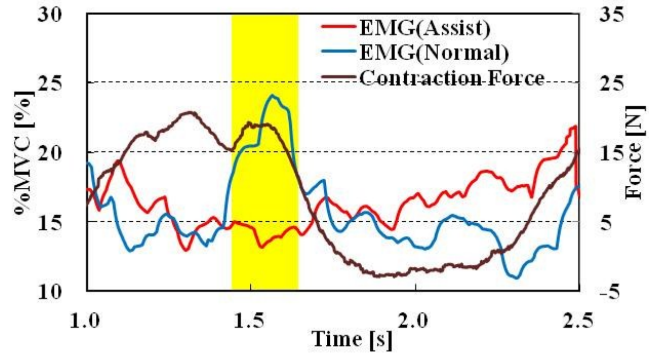

Fig. 13. Sartorius muscle's EMG.

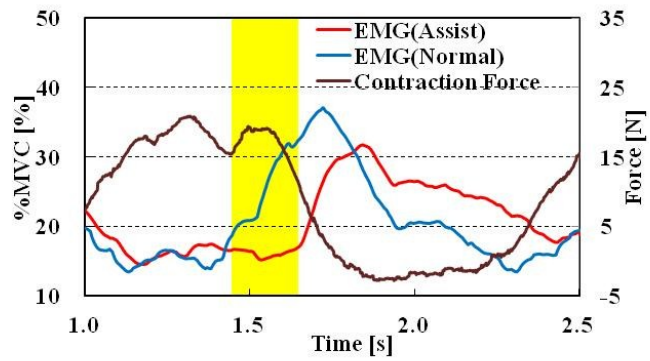

Fig. 14. Semitendinosus muscle' EMG.

C. Consideration

Table II shows that step length is increased by approximately 23 mm as a result of assistance. Fig. 10 shows that displacement in the x-direction on the sagittal plane is increased in the case of assistance. Moreover, since the half of the graph, it turns out that the phase in the case of assistance is delay. It is thought that this is because the walking cycle becomes long due to the increase in step length achieved by assistance.

Fig. 11 shows that the tensile force is produced when the pressure is high. Moreover, Fig. 12 shows that this tensile force rises immediately after the maximum extension of the hip joint; such a force is called "assist-force." However, this tensile force also rises before this point, and such a force is called "nonassist-force." As this tensile force has risen before the maximum extension of the hip joint, there is a possibility of disturbance in the movement of kicking down onto the ground. The nonassist-force occurs because the artificial muscle is strained under the condition of maximum extension of the hip joint, so that the artificial muscle—in which the amount of contraction is low—provides the assist-force efficiently to the leg. Even if the artificial muscle is fixed in such a manner, this occurs at a slack at 5% of the walking cycle after the maximum extension of the hip joint. If the contraction of an artificial muscle starts until this slack is taken up, the assist-force does not occur. Therefore, the assist-force is at approximately 19 N, which is lesser than the desired 40 N. A future task is to develop a new mechanism to solve the decrease in the assist-force and the problem of generating the nonassist-force by the slack of the artificial muscle.

Fig. 13 shows that reduction in EMG can be checked in the area where the assist-force is produced. On the other hand, EMG with assistance after swinging the leg forward is higher than EMG with normal walking. It is thought that to maintain the body balance after an increase in step length, muscular activity is increased.

Fig. 14 shows that a reduction in EMG can be checked in the area where the assist-force is produced. It is thought that this is because of the escape of the entire leg strength due to assistance, causing a decrease in EMG. This also explains why EMG increases as assistive power is lost. Near the local maximum, the leg is swung down toward the ground. After that, EMG with assistance becomes higher sequentially. The reason for this is the same as that for the sartorius muscle, in that to maintain body balance after an increase in step length, the activity of all muscles is increased. Although there was an increase in EMG for maintaining body balance in the stance phase, a reduction of the maximum of EMG could be checked, and the amplitude of EMG became narrow. Therefore, it is thought that a person with little muscular strength can walk like a healthy person. In other words, in the case of this healthy subject, the increase in step length is only 23 mm, but it is thought that further increase in step length can be expected in the case of a person with a slight gait disorder having smaller step lengths from the start.

V. CONCLUSION

A. Result

The increase in step length achieved by assistance from the developed orthosis was evaluated using a healthy subject.

Although there was an increase in EMG as a result of maintaining body balance during the stance phase, it was ascertained that the EMG of the sartorius muscle, which swings the leg forward, decreased and that the strength of the semitendinosus muscle, which pulls the leg back against over-assistance, did not increase but decreased.

It was determined that it is necessary to increase the assist-force and prevent the nonassist-force at the maximum extension of the hip joint.

B. Prospects for the Future

The following is the subject's subjective comment: "I can feel an assistive force, however, small. The weight of the orthosis is worrisome. There is some sense of incongruity at the time of maximum extension of the hip joint."

As a future task, we will develop a new mechanism that cancels out the slack of an artificial muscle without disturbing walking at the time of maximum extension.

We will improve the potentiometer-sensing accuracy so that it can consistently provide appropriate assistance at each step.

We will conduct the experiments with elderly people.

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