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A Walking Assistive Device with Intention Detection using Back-driven Pneumatic Artificial Muscles

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***Abstract*—In this paper, a pneumatically-driven walking assistive device is proposed. Since the structure of the exoskeleton is different from users' legs, it is not required to attach the exoskeleton links to the user's knee and adjust the length of the links. McKibben type pneumatic artificial muscles are adopted as the actuators, realizing back-drivability. We proposed a control method with the detection of the walking intention of the user using air pressure sensors isolated from the device itself, without attaching sensors to the user. The actuators are pre-pressurized and then the difference of the pressure is monitored and used as the trigger to starting the assist. Experimental results show the effectiveness of the proposed system and the intention detection method.**

I. INTRODUCTION

Power assist devices are desired in various fields. They have been adopted in factories to assist the workers lifting and carrying heavy mechanical parts. Together with the evolution of actuator and sensor technologies, wearable power assist systems called powered exoskeletons are developed[1]. The applications of power assist systems are extending to welfare including nursing care and rehabilitation, agriculture, and construction sites.

Various mechanisms and control methods of power assist systems are developed depending on their applications. HAL[2][3][4] and ReWalk[5] are the lower limb exoskeletons for rehabilitation. HAL is driven by electric motors and assists the hip and knee joints. It detects the walking intention of the user by bioelectrical sensors attached to the user. Kilicarslan et. al. developed an intention detection method by probabilistic approach from the electroencephalography (EEG) signals[6]. On the other hand, several systems uses mechanical sensors, without bionic sensors. ReWalk uses motion sensors which detect the changes in center of gravity. Shen et. al. used potentiometer for the joint angles and force sensors to measure the ground reaction force and finite state machine is implemented[7]. "Soft Exosuit" developed in Harvard University[8][9] are lightweight systems without rigid exoskeleton driven by Bowden cables. A walking assist system can prevent the user from falling down by additional control method. Hayashi et. al. proposed a ZMP-based control, which modifies the exoskeleton trajectory using environment sensing[10].

Several exoskeletons can assist the user lifting some loads. Kobayashi et. al. proposed a "muscle suit" using pneumatic artificial muscles as actuators for caregivers [11]. Sano et. al. developed the WAS-LiBERo [12], which is the electrically-driven exoskeleton for transportation of heavy loads in agriculture. Kazerooni et. al. developed Berkeley Lower Extremity Exoskeleton (BLEEX) [13][14]. It is driven by hydraulic actuators and can support up to 75 kg of loads. Pressure sensors are implemented in its foot soles and measure the user's state. Yamamoto et. al. developed an assist device for nurses [15]. The device is driven by pneumatic cuffs and a muscle hardness sensor is proposed to detect the motion intention.

One of the challenges of walking assist devices is the detection of walking intention of the user using while reducing the number of sensors. Many of the exoskeleton uses electromyography (EMG), electroencephalography (EEG), or mechanical sensors such as force sensors for intention detection[16]. The reliability of the intention detection can be increased by fusing multimodal information from different types of sensors. Attaching many sensors to the user, however, requires cost for sensor hardware and time for fixing them.

Another problem is the time to attach the exoskeleton to the user. In conventional exoskeletons, the hip and knee of the user should be fixed to the device using bands. The link length must be adjusted for each user's leg length.

In this paper, a pneumatically-driven walking assist device and its control method are developed. Its link mechanism is different from the human legs and the user only have to fix the hip and the foot without fixing his/her knees or adjusting the link length of the exoskeleton. Pneumatic artificial rubber muscles (PARM) are used as actuators. Since

Fig. 1: The pneumatic lower limb exoskeleton

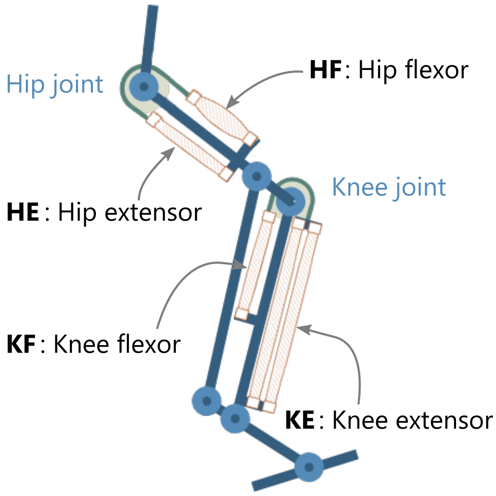


Fig. 2: Mechanism of the device

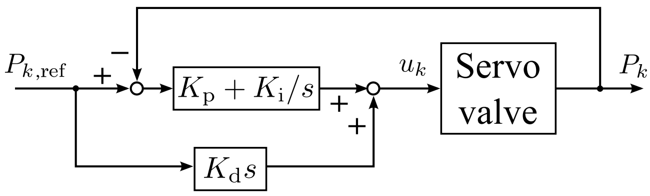


Fig. 3: Pneumatic pressure controller

the PARMs are back-drivable, walking intention can be detected by measuring the change of air pressure in the PARMs. The proposed control method monitors the time derivative of the air pressure of the specified PARM and activates other PARMs when the pressure exceed a threshold.

From the perspective of sensor fusion technology[16] in intention detection, the proposed methods can be a new modality to the existing fusion algorithms. Combining the EMG, EEG, or other signals with the proposed detection method will lead to the increased detection performance or reduced number of sensors.

Experiments are conducted to confirm the effectiveness of the developed system and the results shows that the fatigue of the user can be reduced by the system.

II. System Configuration

The developed system consists of a lower limb exoskeleton and a pneumatic controller.

A. The exoskeleton

Figure 1 shows the prototype exoskeleton and Fig. 2 shows the mechanism of the device. The developed device has a mechanism like a bird's leg, using parallel link mechanism. Since the mechanism is different from that of human legs, the knees are free and its hip parts and foots are fixed to the user using belts.

Two pneumatic artificial rubber muscles (PARMs) antagonistically drive each joint via chains and a pulley. When the PARM contracts, the driving force is transmitted to the pulley fixed to the link. Antagonistic driving is necessary since PARMs cannot apply force in the direction of expansion.

The hip joint and the knee joint is actuated for each leg and the system has four active joints in total. In this paper, each artificial muscle is represented as follows:

KE Knee extensor muscle
KF Knee flexor muscle
HE Hip extensor muscle
HF Hip flexor muscle

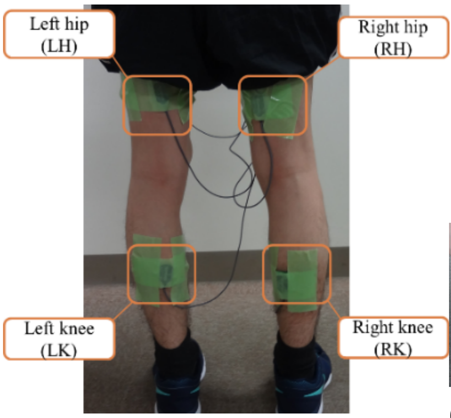
The prefix L and R represent the left leg and the right leg respectively. Each of the KEs has two PARMs and a larger pulley to increase the torque in order to support the heavy shank link.

B. The controller

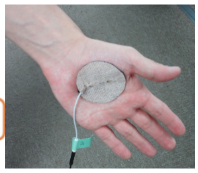
The controller consists of servo valves, air pressure sensors, and a computer. The servo valves are controlled using a computer via a DA converter using the signals of the pressure sensors obtained via an AD converter. The output ports of the valves are connected to the PARMs of the exoskeleton. The controller is separated from the exoskeleton, but we are planning to make it portable in the future.

TABLE I: Parameters of pneumatic force control

Parameter	Value
Proportional gain K_{p}	2.7×10^{-3} V/kPa
Integral gain K_{i}	1.7×10 V/kPas
Differential gain K_{d}	1.0×10^{-4} Vs/kPa



(a) Electrodes



(b) Reference electrode

Fig. 4: Position of EMG electrodes

The pneumatic pressure control is shown in Fig. 3. It is a PI-based control similar to that used in our surgical robot [17]. For a given reference pressure $P_{k,\mathrm{ref}}$, the controller outputs the valve voltage u_k to let the PARM pressure P_k follow $P_{k,\mathrm{ref}}$, where the subscript k denotes the index of the PARM. The PID gains are adjusted manually by checking the step responses and are shown in Table I.

III. Walking Assist Control

Walking assist control should detect the user's intention of walking first, then actuate the appropriate combination of the artificial muscles. This section explains the proposed method of intention detection and the introduction of "base pressure" to increase the sensitivity of the detection.

A. Detection of walking intention

The walking intention of the user can be detected using the back-drivability of the pneumatic actuators. When the user moves his/her leg to start walking, the motion is transmitted to the actuator via the chains and pulleys. If the PARM is pulled and extended, its volume decreases and its internal pressure temporally increases. Contrarily, its pressure increases when pushed. Thus, walking intention can be monitored by measuring the change of the air pressure.

We simply measure the time derivative of the pressure dP_k/dt and activate the PARMs when dP_k/dt exceeds a specific threshold.

B. Base pressure

The chains slack if no air is supplied to PARMs. This slack may be the deadband of intention detection, since the user has to move their leg until the slack disappears.

TABLE II: Experimental conditions

$P_{k,\mathrm{b}}$ [kPa]	100, 110, 120, 130, 140, 150, 160, 170, 180, 190, 200
f_{gait} [steps/min]	0 (Stance), 60, 70, 80
l_{step} [mm]	0 (Stance), 500, 600, 700

In this paper, a base pressure $P_{k,b}$ is applied to the PARMs to pre-tension the chain and detect the walking intention quickly. However, too high base pressure leads to the resistance to the user's motion. The base pressure $P_{k,b}$ are determined experimentally in this section. The following characteristics are considered when determining $P_{k,b}$:

Detectability

Not all PARMs can detect the walking intention with enough sensitivity due to the mechanical and pneumatic characteristics of each joint. Detectability is evaluated using the maximum value of the pressure derivative $\dot{P}_k$, after measuring the pressure by an experiment of walking with a base pressure $P_{k,b}$ but without the assist control. Detectability is defined as follows:

$$D_k = \max_t \frac{dP_k(t)}{dt} \tag{1}$$

Resistance

If the base pressure is too high, the fatigue of the user may be larger. In this work, the user's fatigue caused by the motion resistance of the pressurized actuators is measured by EMG sensors. Note that the EMG sensors are only used for design and evaluation of the control parameters, and they are not used in the assist control. The resistance is evaluated by IEMG defined as follows:

$$\mathrm{IEMG}_k = \frac{\int_0^T \mathrm{NEMG}_k(t)dt}{T} \tag{2}$$

$$\mathrm{NEMG}_k(t) = \frac{\mathrm{EMG}_k(t) - \mathrm{EMG}_k\mathrm{min}}{\mathrm{EMG}_k\mathrm{max} - \mathrm{EMG}_k\mathrm{min}}, \tag{3}$$

where $\mathrm{EMG}_k(t)$ denotes the EMG signal of the muscle k. The subscripts min and max represent the minimum and maximum value of EMG signals. The value T is the time of the experiment. DC offset of the EMG value is removed and then a low-pass filter of 10 Hz cutoff is applied to it before calculating IEMG.

The above criteria are evaluated by experiments with various parameters. In this experiment, the base pressures $P_{k,b}$, the walking frequency f_{gait}, and the walking step l_{step} are varied. The same value of base pressures are applied for each trial: for example, $P_{\mathrm{KE},b} = P_{\mathrm{KF},b} = P_{\mathrm{HE},b} = P_{\mathrm{HF},b} = 150$[kPa]. Electromyographic sensors (Delsys Bagnoli Desktop) are attached to measure the muscle activities as shown in Fig. 4. The tried parameters are shown in Table II.

1) Detectability and base pressure: Figure 5 shows the dependency of the detectability on the base pressure when the user is standing and walking ($f_{\mathrm{gait}} = 70$[step/min] and

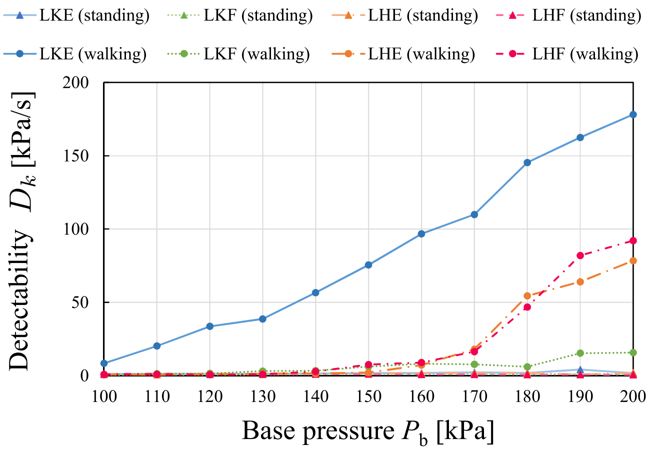


Fig. 5: Relationship between $P_{k,\mathrm{b}}$ and D_k

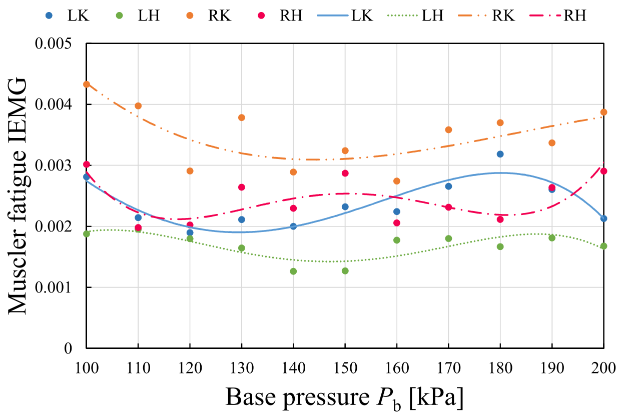


Fig. 6: Relationship between $P_{k,\mathrm{b}}$ and IEMG

$l_{\mathrm{step}} = 600$[mm]). When standing, D_k are almost zeros. On the other hand, D_k becomes larger if $P_{k,b}$ is higher when walking. Thus it is confirmed experimentally that the higher base pressure achieves the higher detectability.

2) Resistance and base pressure: Figure 6 shows the dependency of the fatigue of the user on the base pressure when the user is walking ($f_{\mathrm{gait}} = 70$[step/min] and $l_{\mathrm{step}} = 600$[mm]). The 4th-order approximations are also shown in Fig. 6, just for reference. Each of EMG_k has a minimum value and the fatigue is larger if $P_{k,b}$ is both low and high. When $P_{k,b}$ are too low, the user has to support the weight of the exoskeleton, which is not actuated enough.

3) Determining the base pressure: From above discussion, we determined the base pressures as follows:

KE KE can detect the walking intention whatever the value of $P_{\mathrm{KE},b}$ is. Thus we set $P_{\mathrm{KE},b} = 140$[kPa] only considering the $\mathrm{IEMG}_{\mathrm{KE}}$.

KF, HE, and HF From Fig. 5, KF, HE, and HF can not detect the intention unless $P_{k,b}$ are higher than specific values. Thus, we choose $P_{\mathrm{KF},b} = 200$[kPa], $P_{\mathrm{HE},b} = 180$[kPa], and $P_{\mathrm{HF},b} = 180$[kPa] considering both detectability and resistance.

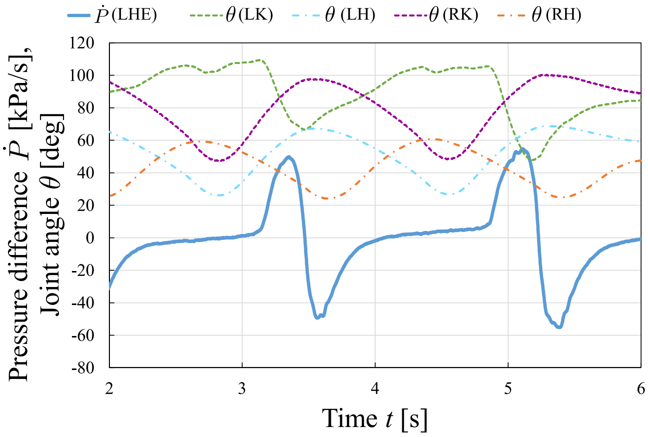


Fig. 7: Joint angles and pressure derivative

C. Detector muscles and active muscles

Next, we determine which of the artificial muscles to be used as intention detectors or actuators. As an example, the pressure derivative of the LHE muscle and joint angles are shown in Fig. 7. When $\dot{P}_{\mathrm{LHE}}$ is positive, i.e. the user's left leg is swinging, the left hip joint is flexed and the left knee joint is extended. As for the right leg, the hip joint is extended and the knee joint is flexed. When $\dot{P}_{\mathrm{LHE}}$ is negative, on the other hand, the direction of motion of each joint is changing. For example, the left hip is changing to flexion to extension.

In this case, the relationship between the sign of $\dot{P}_{\mathrm{LHE}}$ and the motion of each joint is clear, thus $\dot{P}_{\mathrm{LHE}}$ can be used as the trigger to start the assist control.

This detectability is validated in the same way and Table III shows the results for each muscle and joint. A mark "✓" means that the activity of the human muscle causes enough amount of pressure derivative on each artificial muscle. For example, pressure of LHF artificial muscle (4th row) P_{LHF} changes when the user moves any of his/her LKE, LKF, LHF, RKF, or RHE. A muscle without a check cannot detect the intention due to the following reasons:

- The S/N ratio of $\dot{P}_k$ is too low in practical use.
- Multiple motions show the same pattern of $\dot{P}_k$.

Since LHE and RHE can detect all muscle activities as shown in Table III, we determine them as the detector muscles.

D. Control sequence

The developed control method first detects which leg is swinging and then apply the specified pressure to each artificial muscle.

The control algorithm is as follows:

TABLE III: Detectability of muscle activity

		Sensed human muscle							
		LKE	LKF	LHE	LHF	RKE	RKF	RHE	RHF
Sensing artificial muscle	LKE								
	LKF								
	LHE	✓	✓	✓	✓	✓	✓	✓	✓
	LHF	✓	✓		✓		✓	✓	
	RKE								
	RKF		✓	✓	✓	✓	✓	✓	✓
	RHE	✓	✓	✓	✓	✓	✓	✓	✓
	RHF		✓	✓		✓	✓		✓

Initial state
Set $P_{k,\mathrm{ref}}$ to the base pressure and wait for the user input.

Left assist: Phase 1
When $\dot{P}_{\mathrm{LHE}}$ exceeds the threshold (8.7 kPa/s), the system assists the swing of the left leg while assisting the stance of the right leg. The reference air pressures $P_{k,\mathrm{ref}}$ are chosen so that both joints of the left leg are flexed and those of right leg are extended.

Left assist: Phase 2
A specified time (0.25 sec) after, the controller changes the air pressure to assist the left leg standing and the right leg swinging. If the system does not detect the user intention 1.5 sec after that, go to initial state.

Right assist
The algorithm is the same as left assist.

The above algorithm is shown in Fig. 8 and the time-domain sequence of the pressure values is also shown in Fig. 9.

IV. Experiments

A. Experimental conditions

Experiments are conducted to confirm the effectiveness of the proposed assist device and the control system. The users wear the exoskeleton and walk along a straight path of 3000mm with the step length $l_{\mathrm{step}} = 600$mm and the frequency $f_{\mathrm{gait}} = 70$step/min. A metronome is used for the users to keep the exact frequency.

The IEMG values defined in Section III-B are evaluated. In this section, the DC offset of EMG value is filtered out before integration by simply calculate the mean value. For comparison, the user also walks with the EMG sensors without wearing the exoskeleton. A healthy male aged in his 20's participated the experiment.

B. Results and discussions

The experimental results are shown in Fig. 10. Three trials for each experimental conditions are conducted and IEMG of each trial are plotted. The IEMG values is reduced

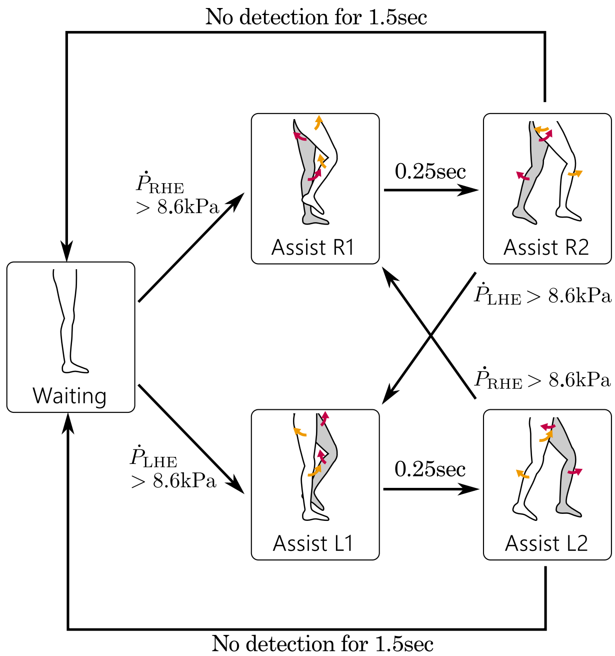


Fig. 8: State diagram of assist control

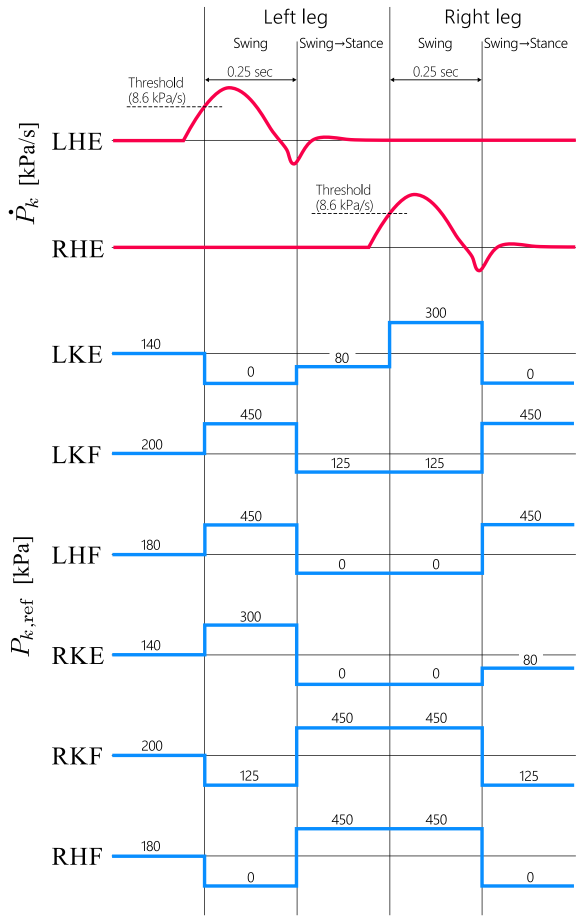


Fig. 9: Sequence of assist control

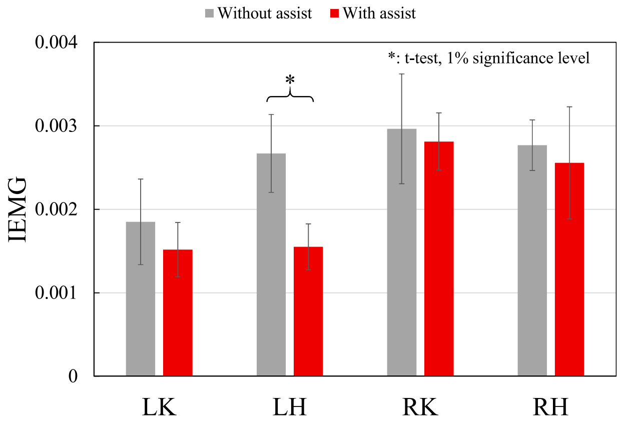


Fig. 10: Experimental result

when the assist device is worn and activated. The IEMG of the LH joint is significantly lower ($p = 0.01$). The effectiveness of the exoskeleton system is validated, but a more lightweight mechanism is desired since the output of the artificial muscles is used to support the weight of the mechanism in large part.

The performance of intention detection is high in this experiment, but it is expected that failure of detection frequently occurs if the device is tried by more users. Detection accuracy can be improved by introducing secondary sensors such as pressure sensors in the sole of foot. Foot pressure can be measured by introducing a pneumatic balloon, which can be easily implemented in our pneumatic control box. It is also necessary, in contrast, to test whether the performance of existing intention detection methods can be increased by combining the proposed method.

V. Conclusion

In this paper, a lower limb exoskeleton using pneumatic artificial rubber muscles is developed. Its mechanism is different from the human legs and it is not necessary to adjust the link length of the device depending on the user's leg length. An assist control method without attaching EMG sensors or force sensors to the user, detecting the walking intention by monitoring the derivative of air pressure, is proposed. The criteria to design a controller, intention detectability and resistance, are introduced. Experimental results show the significant reduction of the user's fatigue for some of the muscles. Future works are to develop a more lightweight mechanism with enough output force and a portable pneumatic control system.

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