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PVC gel soft actuator-based wearable assist wear for hip joint support during walking

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Abstract

Plasticized polyvinyl chloride (PVC) gel and mesh electrode-based soft actuators have considerable potential to provide new types of artificial muscle, exhibiting similar responsiveness to biological muscle in air, $>10\%$ deformation, >90 kPa output stress, variable stiffness, long cycle life (>5 million cycles), and low power consumption. We have designed and fabricated a prototype of walking assist wear using the PVC gel actuator in previous study. The system has several advantages compared with traditional motor-based exoskeletons, including lower weight and power consumption, and no requirement for rigid external structures that constrain the wearer's joints. In this study, we designed and established a control and power system to making the whole system portable and wearable outdoors. And we designed two control strategies based on the characteristics of the assist wear and the biological kinematics. In a preliminary experimental evaluation, a hemiparetic stroke patient performed a 10 m to-and-fro straight line walking task with and without assist wear on the affected side. We found that the assist wear enabled natural movement, increasing step length and decreasing muscular activity during straight line walking. We demonstrated that the assistance effect could be adjusted by controlling the on-off time of the PVC gel soft actuators. The results show the effectiveness of the proposed system and suggest the feasibility of PVC gel soft actuators for developing practical soft wearable assistive devices, informing the development of future wearable robots and the other soft actuator technologies for human movement assistance and rehabilitation.

Keywords: PVC gel actuator, soft actuator, artificial muscle, electroactive polymer, walking assist wear, application

(Some figures may appear in colour only in the online journal)

1. Introduction

With a rapidly aging society, an increasing number of elderly people require care after suffering from stroke and other age-related disabilities. Various technologies, devices and robots are emerging to aid caretakers in the physically and emotionally exhausting work of care for older people. One of the most interest area is the development of devices to assist the lower body in tasks such as walking and supporting heavy loads. Walking is an important daily activity, providing physical exercise and enabling shopping and outdoor social activities. Aging can cause reduced mobility, muscular atrophy, gait disorder and even falling-related accidents that can lead to death [1–3].

In the last several decades, a range of orthoses and exoskeletons for human augmentation, rehabilitation and medical assistance have been developed, and some have been commercialized. The vast majority of these systems involve motors and rigid frame-based exoskeletons which can apply torque to the joints and support compressive forces. Some systems have been developed to help carry heavy loads, decreasing the wearers' metabolic cost [4–6]. Other devices

[3] Member, IEEE.

are designed to supply additional energy for movement in daily life, for able-bodied individuals, older people, and people suffering from lower-limb disabilities [7–12]. These systems have the advantage of providing sufficient bandwidth and torque for augmenting the performance of the wearer during walking. However, exoskeletons still present challenges for widespread use, including difficulty fitting the equipment [13], unexpected and potentially dangerous internal joint forces due to misalignment [14, 15], and the need for complex and bulky self-aligning mechanisms to solve these problems [15–18]. In addition, although the weight of an exoskeleton can be supported by itself, the inertia can still be felt by the user. Thus, these systems tend to restrict some natural movements during daily life, including external and internal rotations of the hip joint, placing a burden to the wearer [19].

Soft robotic assist wear has a number of advantages over rigid exoskeleton devices. They are flexible, compliant, lightweight and easy to fit to an individual, and place fewer limitations on the natural movements of the wearers' joints. However, these systems involve challenges in transferring power from the body to the ground without rigid frames, and the complex biomechanics of human body can cause difficulty in mounting the actuators and sensors. To date, most soft robotic assist wear devices have been constructed using pneumatic muscle actuators (PMAs), which have the advantages of high force to weight ratio, light weight, low cost, and flexibility. Costa *et al* [20] developed an orthosis using pairs of PMAs in parallel to directly provide torque on the joint through pulleys. Other PMA-based systems have directly used the tensile force of PMAs to create torque through the biomechanical structure of human body [21–23]. These systems have been shown to be effective through different innovative design features and control strategies [24]. However, the need for a heavy pneumatic power supply system with a limited continuous working time and substantial noise present challenges for practical use. Besides, several different soft wearable assist devices have recently been proposed, using soft elastic belts [25] and Bowden cables [26, 27] driven by geared motors. Because these devices are driven by motors arranged on the waist or back of the wearer, the motors and controllers require substantial power and weight. Therefore, new actuator-based technologies necessary for overcoming the challenges of current wearable assist wear for use in daily life.

Polymer-based soft actuators are lightweight and flexible, similar to human muscular actuators, and have attracted substantial attention in the last several years [28, 29]. However, few soft actuators based human motion assistive devices have been developed so far due to different difficulties of actuators, such as limitation of driving in solution [30], requirement of high driving voltage or temperature [31, 32] and with slow response rate and high-power consumption [33]. We have developed a soft actuator using PVC gels and stainless mesh electrodes [34]. It shows substantial potential for practical use due to a range of features, including: (1) the actuator is lightweight, flexible, low-cost and easy to fabricate, (2) the actuator exhibits stable performance under conditions of substantial displacement and notable output stress in air, (3) a driving electric field of low strength ($\sim$2 V/um) compared with dielectric elastomer actuators ($\sim$418 V/um) [35], (4) low power consumption ($\sim$5 μW/um) and a long cycle life (over 5 million cycles). Our long-term goal is to develop a lightweight, portable and wearable assist wear device using PVC gel soft actuators that can be used in daily life to help the wearer to reduce muscular burden compared with regular walking. We recently proposed an initial version of an assistive device using the stiffness variation of PVC gel soft actuators [36]. Although we experimentally verified the feasibility of using the variable stiffness of PVC gel soft actuators for motion assistance, the system has a weak structure that can easily break. Thus, we proposed an assistance device with a new structure and design, using a contraction-expansion output force for assisting motion that provided better performance and was more robust in relation to external forces [37]. The system has a number of advantages, including its light weight, compact and flexible structure, ease of fitting, ease of attachment and removal. We have designed and fabricated a prototype of the new-type assist wear. However, the control and power system is not established for a practical use, and proper control strategy and methodology should be given and the effectiveness of the new-type assist wear remains to be validated.

In the current study, we have designed and established the control and power system of the new-type assist wear, making the whole system portable and wearable outdoors. And we designed two control strategies based on the characteristics of the assist wear and the biological kinematics. Finally, we conducted a preliminary experiment to evaluate the effectiveness of the proposed system in terms of physiology, physics and psychology, to demonstrate and showed the feasibility and practical use of the proposed system.

2. System overview

Muscular strength decreases with age, limiting the joint range of motion and causing loss of balance that can result in falling during walking [38, 39]. The current study sought to develop a lightweight soft wearable assist wear for supporting activities of daily life for older people with weakened muscles and those with mobility problems.

In this preliminary study, we focused on motion support of the hip joint, because it plays an important role in walking. As shown in figure 1(a), when the hip joint controls the balance and makes an accelerated forward movement to propel the body weight forward, both the moment and power at the hip joint reach peak values within 50%–62% of the walking gait cycle during level-ground walking at normal speed [40]. We think that during this phase of the gait cycle, the PVC gel soft actuator-based assist wear may add an external tensile assistive force that is offset from the hip joint center, potentially creating a torque around the joint to decrease the burden on the leg muscles and increase the step length (figure 1(b)).

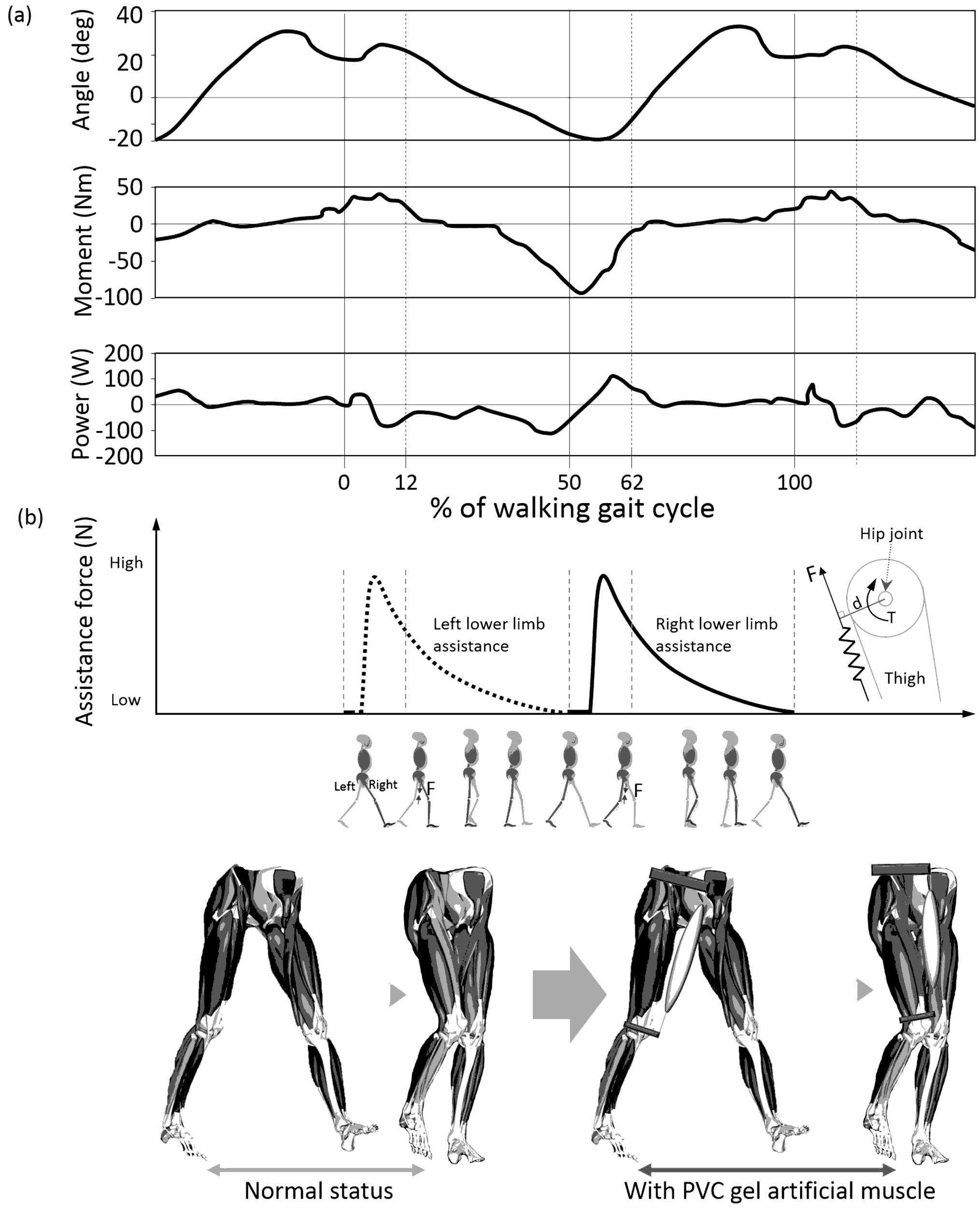

Figure 1. (a) Representative angles, moments, and power of the leg hip joint over the gait cycle. Data are adapted from [40]. (b) Extra assistive force variation and assistive effect expectation.

2.1. Composition of the assist wear system

Figure 2 shows an overview of the prototype PVC gel soft actuator-based walking assist wear system. The device primarily consists of three components: an actuation mechanism to provide tensile force, an insole force sensor to detect gait, and a portable battery and controller to provide power and appropriate assistance during walking. In the actuation component, two modules of PVC gel soft actuators are arranged along the top of the thigh using structured textile belts to transmit the forces across the body to create torque on the hip joint using the contraction-expansion movement (see figure 2(c)). The gait sensing component detects the gait cycle based on changes in the contact area between the foot and ground, which is estimated by the force changes of the insole force sensors (see figure 2(d)). The controller component operates the assist wear according to the status of the walking motion (see figure 2(e)). Using this structure, the whole device can be flexible and avoid constraining the natural movement of the legs, such as abduction/adduction and external rotation/internal rotation. In addition, the system fits easily to the body shape, unlike motor-based exoskeletons that need to precisely align the axis of the motor with the wearer's joints. Moreover, the system can easily adapt to individual height differences by adjusting the length of the textile belts.

The device is designed to assist the lower limb from the phase of pre-swing (the heel strikes the ground) to terminal swing (the heel strikes the ground) (see figure 1(b)). During the swing phase, the controller changes the applied DC

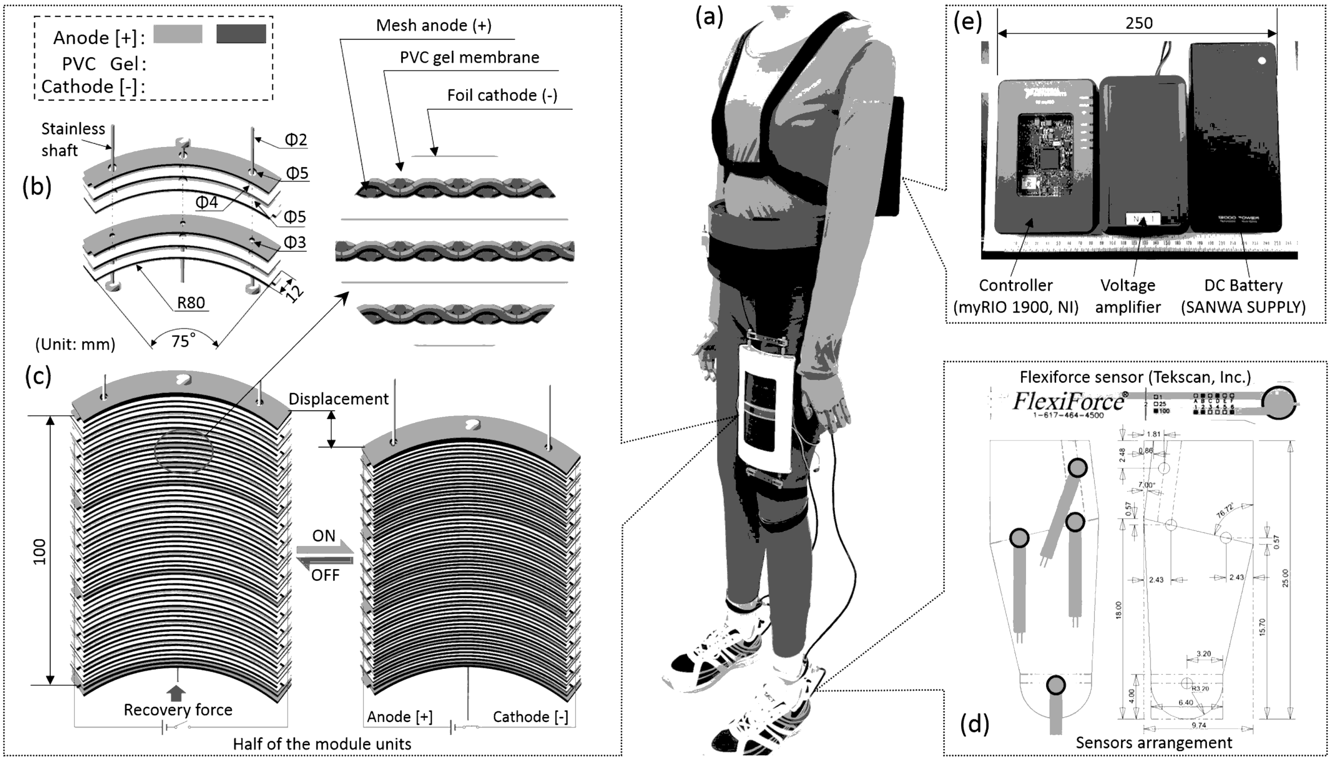

Figure 2. Overview of the prototype of the PVC gel soft actuator-based walking assist wear (a) Overview of wearing set-up of the assist wear. (b) Structure of the multilayered PVC gel actuator with two types of anode mesh electrodes. The red layer with small holes is comprised of slide electrodes to minimize the friction with the slide shafts. (c) Contraction and expansion movement of the stretching type actuator with the DC field turned on and off. (d) FlexiForce sensor-based motion detection (position estimator). (e) Power and controller.

voltage to cause a contraction movement in the device, providing a tensile force to support hip flexion. When the heel contacts the ground at the terminal swing phase, the applied DC voltage changes to produce an extension movement in the device, decreasing the output force to enable easy hip extension of the limb.

2.2. PVC gel soft actuator for actuation

PVC gel deforms asymmetrically between the anode and cathode electrodes under electrical stimuli. A creep deformation happens only on the surface of the anode electrode due to an accumulation of negative charges in the gel near the anode surface. There is no such deformation on the cathode electrode. Therefore, we use a stainless mesh electrode as the anode to enlarge the deformation, and stainless foil as the cathode to minimize the weight of PVC gel soft actuators (see figure 2(c)). When the DC field is turned on, a creep deformation of PVC gel happens along the surface of the anode, causing the gel to move into the holes of the mesh. This brings a contraction deformation of the actuator in the thickness direction. When the DC field is turned off, the actuator returns to its original shape rapidly because of the material elasticity of PVC gels.

When the applied DC voltage is 400 V ($2\,\mathrm{V}/\mu\mathrm{m}$), the PVC gel soft actuator has a contraction strain of about 10%, an output stress of about 90 kPa and a response rate of 9 Hz under a power consumption of $2.4\,\mu\mathrm{W/mm^2}$·layer [37]. Besides, it was confirmed that the PVC gel soft actuators' cycle life is more than 5 million times given a continuous electric field driven at 2 Hz [34]. These results indicate the feasibility of applying the PVC gel soft actuators to practical applications.

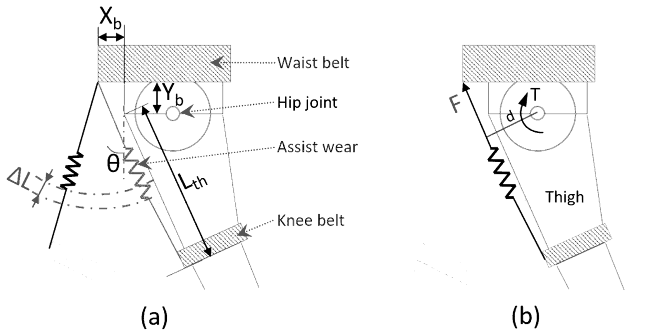

Figure 3. (a) The length of the assist wear varied during the walking gait cycle in the sagittal plane. (b) A moment (T) on the hip joint is created by the tensile force (F) which is shift from the center of the hip joint.

For an effective assistance, it is necessary to determine a range of specifications, such as the maximum displacement, output force and robust structure.

The displacement of the actuator was designed to satisfy the length change of the assist wear from the pre-swing phase to the terminal swing phase, to avoid limiting the range of hip joint motion (figure 3(a)). We found that the length change ΔL was determined by the thickness of the edge profile of the waist belt X_b, and that ΔL increased with an increase of X_b.

Table 1. Characteristics of the PVC gel soft actuators.

	PVC gel	Anode	Cathode
Material	PVC:DBA = 1:4	Stainless mesh	Stainless foil
Size/shape	A curve shape 120 mm in long side 100 mm in short side	#100 (100 wires per inch) The same shape as PVC gels	The same shape as PVC gels
Thickness	about 0.2 mm	0.18 mm (wire diameter of 0.09 mm)	0.01 mm

ΔL can be obtained by the following equation:

$$\Delta L = \sqrt{(L_{\rm th}\sin\theta - X_b)^2 + (L_{\rm th}\cos\theta + Y_b)^2} - \sqrt{(L_{\rm th}\sin\theta_0 - X_b)^2 + (L_{\rm th}\cos\theta_0 + Y_b)^2} \quad (1)$$

Where, the $\boldsymbol{L_{\rm th}}$ is the length of thigh, Y_b is the height between greater trochanter and waist belt, $\boldsymbol{\theta}$ is the angle of thigh during walking, $\boldsymbol{\theta_0}$ is the angle of thigh at the end of terminal swing phase.

Based on the average size of the human upper leg, we obtained a length change (ΔL) of approximately 13.5 mm when X_b was set to 20 mm [37]. The displacement of the PVC gel soft actuator was almost linear in proportion to the number of stacked layers. Considering the average length of the human leg, we set the displacement of the assist wear to approximately 20 mm, which is greater than the length change of the assist wear. When the applied DC voltage was 400 V, the height of the total artificial muscle was approximately 200 mm according to a 10% contraction strain of the actuator.

To determine the output force, as shown in figure 1(a) the moment on the hip joint reached a peak value at the beginning of the flexion motion, and the maximum moment at the hip joint was approximately 70 Nm (i.e., a weight of approximately 70 kg) [41, 42]. Our intention was to develop a system for long-term walking support during daily life rather than a power assist suit, we thought that the assistance could be effective with a relatively small torque support for a long time use. To make the assist wear be compact and lightweight so as not to give a burden to the wearer, we aimed for approximately 10% (7 Nm) assistance of the maximum torque on the hip for the proposed prototype. As shown in figure 3(b), the torque arm was approximately 0.1 m, in accord with the average size of human body [43, 44]. Thus, we were able to obtain the desired force of approximately 70 N. The output force of the PVC gel soft actuator was almost linear in proportion to the area of the single-layer structure actuator. We set the maximum output force of the actuator to approximately 80 N, which is greater than the desired force, and we calculated the desired area of the actuator as approximately 0.001 m^2 when the DC voltage was 400 V. The PVC gel in this study is constructed from commercial PVC powder (with the degree of polymerization of approximately 3200) plasticized with dibutyl adipate (DBA) plasticizer. The characteristics of the PVC gel actuator can be changed by adjusting the weight ratio of PVC and DBA, and the mesh size of the anode. Table 1 shows the characteristics of the PVC gel soft actuator in the current study.

To avoid breakages in the stacked single layers of the actuator, we used a stretching-type structure comprising two slide frames. As shown in figure 2(c), when the DC voltage was turned on, the PVC gel soft actuator contracted and led the middle slide frame downward. When the voltage was turned off, the actuator recovered to its original shape and brought the middle slide frame upward to generate a tensile force to support the motion at an appropriate time. We divided the whole PVC gel soft actuator into two stretching-type modular units to construct the assist wear. Furthermore, to ensure appropriate fitting to the human body, we designed the stretching-type actuators with a curved shape according to the average size of the older human body [43, 44]. For safety, we used a 0.5 mm thick Teflon cover which can withstand a voltage of over 2 kV to ensure insulation and protection.

Since the process of manual fabrication of the multi-layered PVC gel soft actuators is time consuming, we fabricated the assist wear for only one side in this preliminary study. The fabricated assist wear had a total weight of about 0.6 kg (for one leg), including accessories (frame of mechanical stretching-type structure, Teflon cover, belt etc), as shown in figure 2(a).

2.3. Sensor system

Traditional motor-based rigid exoskeletons usually use sensors such as encoders or potentiometers in robotic joints to track joint angles. However, these sensing technologies are not compatible with our soft assist wear. In this study, we used flexible FlexiForce® force sensors (Tekscan, Inc.), which are ultra-thin, lightweight and cost effective, offering easy integration into the insoles to measure gait status.

We designed an electrical circuit using a resistance-shunt method in which the sensor acts as a force-sensing resistor [37]. The resistance of the force sensor is high (up to several tens of MΩ) when unloaded, and can be dramatically reduced to several kΩ when a force is applied to the sensor. We arranged four sensors on each foot to construct the insole sensor system, based on the skeletal structure [43] of the foot. Figure 2(d) shows the precise dimensions of the arrangement of sensors for an average male foot size. The sensors were fixed between two natural rubber sheets, making the insole flexible for comfortable walking.

Figure 4 shows an illustration of the right lower limb assistance according to the contact area changes and force sensor state changes during a walking gait cycle. The moment at which the wearer raises the right leg to make a hip flexion from the pre-swing phase to the initial swing phase was detected when the force of the sensor positioned at the big toe (S5) reached a minimal value, and the forces of the sensors (S1-S4) on the opposite foot exceeded a minimal value. By

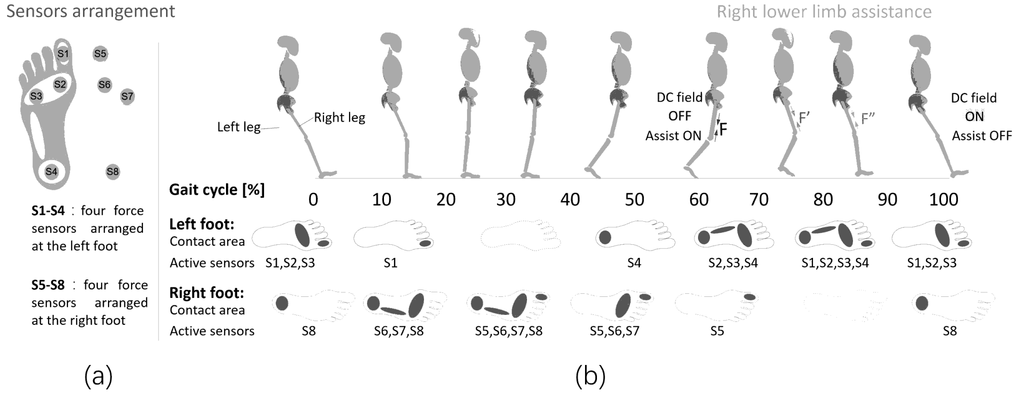

Figure 4. (a) Sensors arrangement on the feet. (b) Contact area variation between the foot and the ground and active force sensor variation during a walking gait cycle.

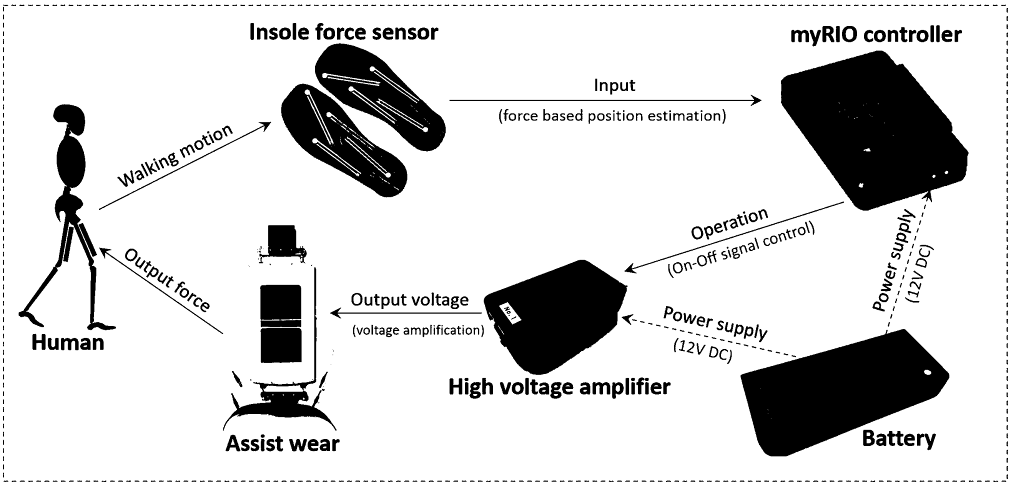

Figure 5. Control structure of the PVC gel soft actuator-based walking assist wear system.

combining the conditions of the status of each sensor on both sides of the foot, we can achieve more accurate and robust detection and estimation of the walking gait cycle, to control the assist wear effectively.

2.4. Control system

Figure 5 shows the control structure of the simple open-loop control system. The control system consisted of a LabVIEW device (myRIO, NI), a high voltage amplifier and a 12 V DC battery (700-BTL012BK, SANWA SUPPLY INC.). To enable real time control according to the walking gait, we used the myRIO device to receive and convert data from the force sensors to estimate walking status and provide appropriate output DC voltage signals to the actuators, to adjust the assistive force. The total weight of the controller is approximately 0.8 kg, including the battery (0.35 kg), the case and belts. The entire weight of the whole assist wear system is about 1.4 kg (2 kg if for two legs ($0.6 \times 2 + 0.8$ kg)).

2.5. System performance

The system performance was evaluated in terms of displacement, output force, response time and power consumption of the assist wear and insole force sensor-based walking gait cycle detection and output control [37].

The displacement of the assistive device was approximately 16 mm (strain of 8%) under an applied voltage of 400 V. Although this was less than the desired displacement of 20 mm, it is greater than the length change of 13.5 mm from the pre-swing phase to the terminal swing phase which would be sufficient for assistance during walking. The maximum output force was approximately 94 N, which was greater than the desired force, ensuring more than 10% assistance of the maximum moment on the hip joint. When the DC field is turned off, the output force decreases nonlinearly with the displacement decreases, and reaches to zero when the actuator returns to its original shape. Equation (2) shows the relation between the displacement (X) and the output force (F).

$$(F + \mathrm{a_f})(X + \mathrm{b_x}) = \mathrm{c} \tag{2}$$

Where, $\mathrm{a_f}$, $\mathrm{b_x}$ and c are constants. $\boldsymbol{F}$ and $\boldsymbol{X}$ can be obtained by the experimental data. $\mathrm{a_f}$, $\mathrm{b_x}$ and c can be obtained by regression analysis method [36]. In this study, $\mathrm{a_f} = 18.9$, $\mathrm{b_x} = 2.9$ and $\mathrm{c} = 330.5$.

In addition, we confirmed that the response time of the assistive component was approximately 56 ms, and the power

Table 2. Comparison of the basic characteristics of different types of soft robotic assist wears.

System		Support joints	Maximum Force at hip joint (N)	Nominal biological torque at hip joint (%)	Total weight (kg)	Power consumption (W)
PVC gel soft actuator based	this work*	hip	94	10	2	3.2
	Li Y *et al** [35]	hip	<10	<2	0.6 (without power and controller)	<2
Pneumatic	Kawamura T *et al* [21]	hip	40	5	4 (without air supply)	—
	Wehner M *et al* [23]	hip, knee, ankle	235	35	7.1 (without air supply)	—
Electro-mechanical	Jin S *et al* [25]	hip	25	<5	2.7	120
	Alan T Asbeck *et al* [26]	hip	150	30	7.57	>100
	Alan T Asbeck *et al* [27]	hip, knee, ankle	150	19	5.5	50

*The total weight and power consumption are calculated for two legs.

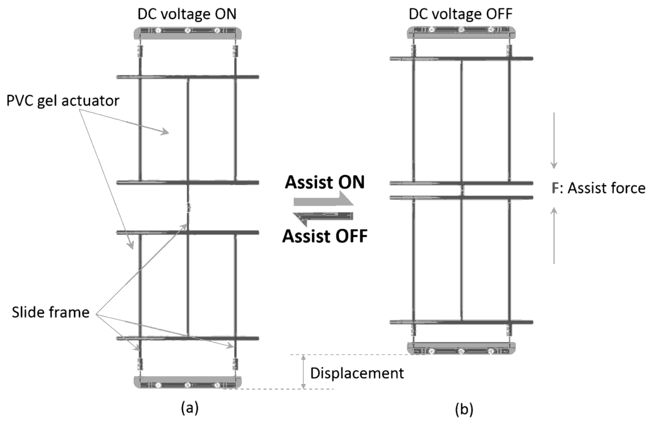

Figure 6. Illustration of the movement of assist wear (front view). (a) Assist OFF. (b) Assist ON.

consumption was approximately 1.6 W. The walking detection and output control component also exhibited good performance during normal walking. The whole system was powered by a 12 V DC battery, which can provide over 10 h continuous driving during walking.

Table 2 shows the comparison of the basic characteristics of different types of soft robotic assist wears. We can see that, the PVC gel soft actuator based assist wear in this work has a relatively light weight and a low power consumption compared with the pneumatic actuators based systems [21, 23] and the electro-mechanical motors based soft robotics devices [25–27].

3. Evaluation of assistive performance

In a preliminary study to evaluate the assistance performance of the proposed PVC gel soft actuator-based walking assist wear, we enrolled a hospitalized hemiparetic stroke patient to participate in an experiment. We selected a hemiparetic patient because the assist wear is designed to support a single leg during walking. We investigated the effects of wearing the assist wear on aspects of physical, physiological and psychological. The patient provided written informed consent to participate in the study, and the experiment was approved by the ethics committees of the university and hospital, respectively.

3.1. Purpose and hypotheses

Hemiparesis, or one-sided weakness in the leg, can cause a loss of balance, difficulty walking due to limited step length and frequency [45]. In the current experiment, we tested a prototype assist wear device on the weakened leg of a patient with hemiparesis to evaluate its effectiveness during steady-state level ground walking. As shown in figure 1(b) the system operates as an external artificial muscle, connected in parallel with the biological muscles around the hip joint. Therefore, we hypothesized that the assistance of an artificial muscle would have positive effect on the walking ability of the weakened leg during activities of daily life and rehabilitation exercises, such as increasing walking speed and step length, and decreasing the level of muscular activity during walking.

3.2. Control algorithm and assistance force estimation

In this system, the moment the wearer raises the foot from the pre-swing phase to the initial swing phase can be detected by the value of the insole force sensor positioned at the big toe reaching a minimal value. The applied electric field is then turned off (the actuator extends to its original shape) to create a contraction movement in the assist wear (Assist ON), to provide a tensile force to support the flexion motion of the lower limb, as shown in figure 6(a). A moment is then created around the hip joint to assist the flexion motion, due to the offset of the tensile force from the hip joint center. The moment that the heel contacts with the ground at the terminal swing phase is detected when the insole force sensor positioned at the heel reaches a threshold value. The applied electric field is turned on (the actuator contracts) to cause an extension movement in the assist wear (Assist OFF), to create slack and release the tensile force for easy extension of the lower limb, as shown in figure 6(b). A combination of the status of the other sensors could ensure accurate estimation of walking gait.

The assistance force of the assist wear changes with the deformation of the PVC gel soft actuators when the DC field is turned off. The force reaches a peak value at the moment of toe off and a minimum value at the time of heel strike. The response time of the soft actuator plays an important role in determining the moment the assist wear transfers the force to the wearer when the DC field is turned off. In this experiment, according to the characteristic of the actuator and biological kinematics, we designed two control strategies (A and B) with different on-off timing to control the system, as shown in figure 7.

Figure 7(a) shows the result of applying a voltage change along with force variation of the right foot during walking, using control strategy A. It can be seen that the applied voltage is turned off (Assist ON) when the value of the sensor positioned at the big toe (S1) reaches a minimal value together with the other three sensors. The applied voltage is turned on (Assist OFF) when the value of the sensor positioned at the heel (S4) exceeds a threshold value. For control strategy B, because the response time of the soft actuator is in the range of several tens of milliseconds (similar to the elapsed time from the peak to the minimum value), we designed another algorithm to turn off the DC field at the peak force value of the sensor positioned at the big toe (see figure 7(b)). The results revealed that the applied voltage was turned off (Assist On) at close to the peak value of the sensor positioned at the big toe (S1), and turned on (Assist Off) when the values of the sensor arranged at the heel (S4) increased from 0. These results are consistent with the desired operation, demonstrating that the force-based position estimation and output control systems functioned appropriately.

Figure 8 shows the estimation of the assistance force and period for control strategies A and B during a walking gait cycle. The average data of the torque and angle (θ) in the

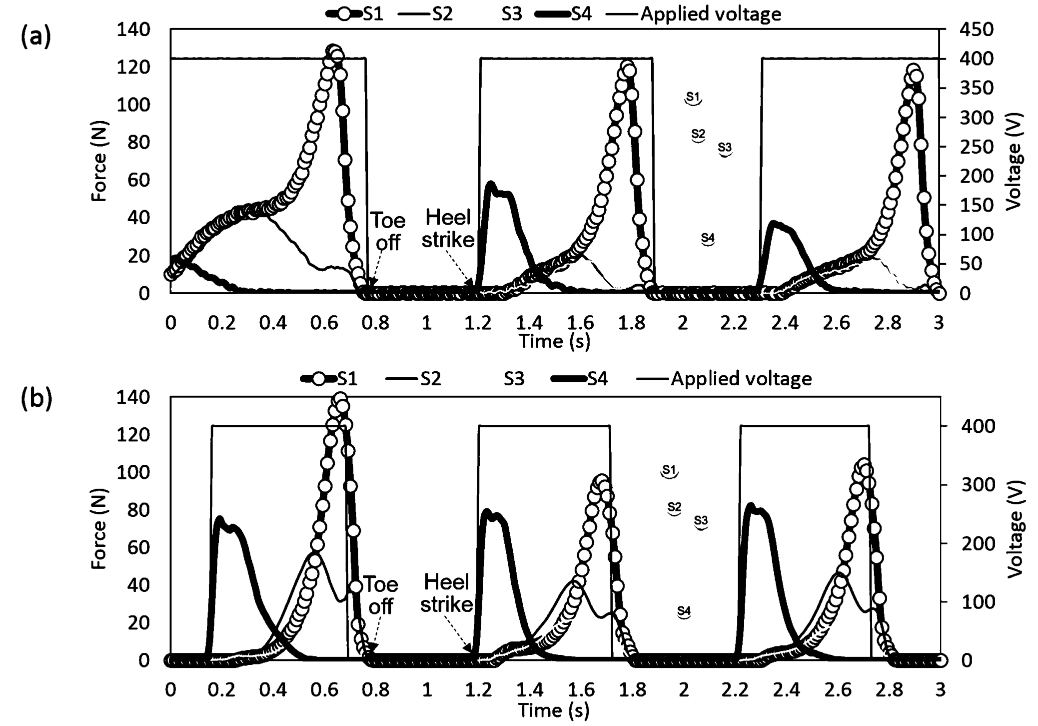

Figure 7. Control strategies and measured results of the walking assist wear. (a) Control strategy A. (b) Control strategy B.

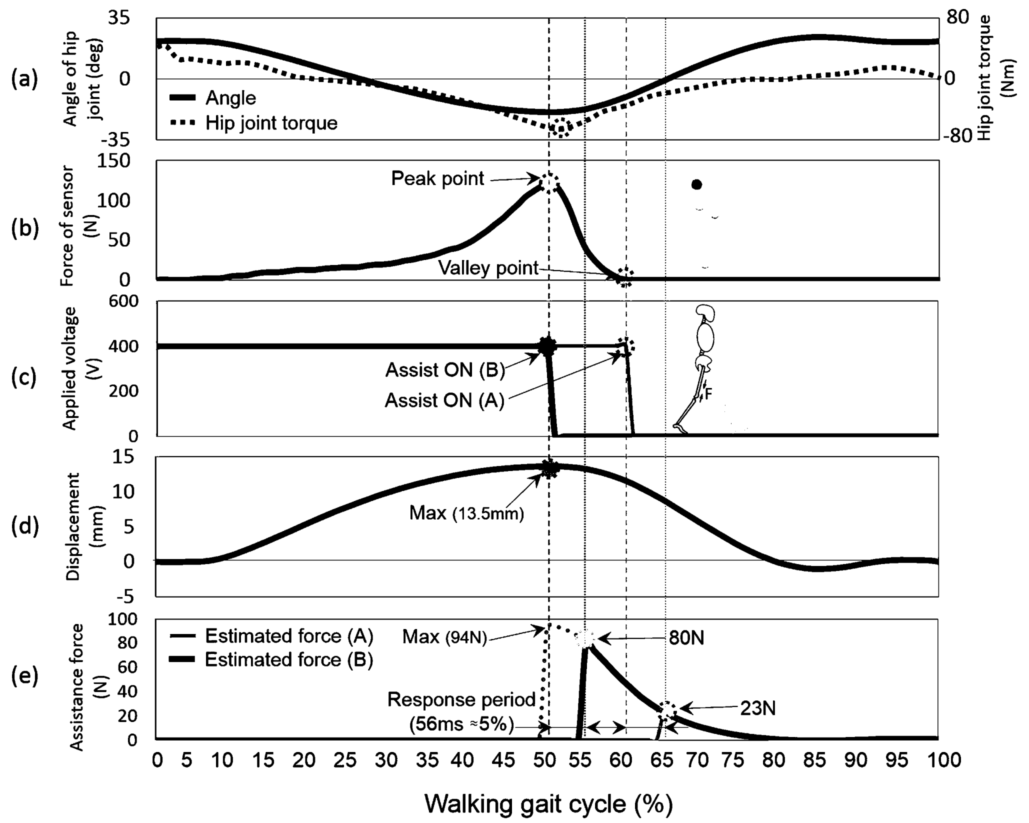

Figure 8. Assistive force estimation for control strategies A and B. (a) Average angle and hip joint torque during a walking gait cycle. (b) Force variation of the sensor positioned at the big toe. (c) Variation of the applied voltage on the soft actuator. (d) Displacement variation of the assist wear during a walking gait cycle. (e) Estimated assistive forces for control strategies A and B.

Table 3. Characteristics of the participant in the walking experiment.

Items	Characteristics
Sex/Age	Male/74 years
Height/Weight	168 cm/67 kg
Foot size	25.5 cm
Affected side/Month since stroke	Right leg/3 months
Walking ability/ Walking speed	Independent (without orthoses)/0.8 m s^{-1}
Degree of paralysis (12-grade recovery grading system of hemiplegia)	11

sagittal plane are quoted from a reference of [46]. And we can obtain the displacement of the assist wear by equation (1) (see figure 8(d)). Then, we can calculate the generation force using equation (2) (see figure 8(e)). When the DC field was turned off, there was a response period that took approximately 5% of the gait cycle before the assist wear transferred the force to the wearer. The output force decreased along with the decrease of displacement from the pre-swing phase. It can be seen that control strategy B produced a stronger maximum assistance force (over 80 N), approximately four times stronger than that of control strategy A. In addition, the assistance period of control strategy B was increased by 10% of a walking gait cycle compared with control strategy A.

The results indicate that control strategy B may generate a greater assistance effect than strategy A. However, an earlier assist force on the thigh may place a burden on the muscles from the terminal stance phase to the pre-swing phase. Thus, we investigated the influence of the two control strategies on walking assistance effectiveness for a hemiparetic stroke patient during a walking task.

3.3. Experimental evaluation

Since the PVC gel soft actuator-based assist wear provides a relatively low level of assistive force compared with traditional motor-based exoskeletons, we tested a hemiparetic stroke patient with a mild stroke, who was hospitalized for 3 months. The patient was able to walk by himself without any orthoses. Table 3 shows the details of the participant in this study.

Figure 9 shows an overview of the experimental design. The participant performed a 10 m to-and-fro straight line walking task with and without the assist wear on the affected (right) side in the hospital. A physician accompanied the participant throughout the experiment for safety.

The experimental procedure was as follows:

(1) The subject was asked to perform the to-and-fro straight line walking task twice, at a comfortable speed, without the assist wear.
(2) The subject was asked to perform the to-and-fro straight line walking task twice, at a comfortable speed, with the assist wear using the control strategy A.
(3) The subject was asked to complete an impression evaluation questionnaire, and to comment freely on their experience of step 2).
(4) The subject was asked to perform the to-and-fro straight line walking task twice at a comfortable speed, with the assist wear using the control strategy B.
(5) The subject was asked to complete an impression evaluation questionnaire, and to comment freely on their experience of step 4).
(6) Again, the subject was asked to perform the to-and-fro straight line walking task twice, at a comfortable speed, without the assist wear.

The subject was not aware of the trial sequence of the two control strategies and was not informed whether or not assistance was provided during the experiment. We measured the walking speed, length of steps, muscular activity (electromyography; EMG), and the acceleration and angular velocity of the upper body during the walking task. We used two digital cameras at the front and the side of the subject to record the walking motion (see figure 9(b)). Walking speed and step length were calculated based on the time and number of steps in one direction, obtained from the videos. As shown in figure 9(c), we investigated muscular activity variation by measuring EMG changes in the rectus femoris muscle, sartorius muscle and hamstring, to determine whether the assist wear reduced the burden on the muscles during walking. In addition, we examined the gastrocnemius with EMG to determine whether the assist wear imposed a burden on the muscle during the stance phase, which could potentially be caused by incorrect attachment or control time of the assist wear. A wireless EMG logger (LP-MS1002, Oisaka Electronic Equipment Ltd, JAPAN) was used to perform EMG measurement. Furthermore, we measured the acceleration and angular velocity variation using two wireless motion recorder (MicroStone Corporation, JAPAN) arranged on the participant's chest and waist, respectively (see figure 9(d)).

4. Results and discussion

4.1. Walking motion detection and output control

We measured the variation of the insole force sensors, the output DC voltage and the current of the soft actuators to evaluate the system during the walking experiment. Figures 10(a) and (b) show the results of the force-based walking motion detection, output voltage control and the current variation of the system during a 10 m one-way walking task, for control strategies A and B, respectively.

As can be seen from the results, the output voltages were correctly turned off (Assist On) to provide an assistive force at the valley point (toe off) and the peak point (heel off) of the S1 for strategies A and B, respectively. In addition, the output voltages were turned on (Assist Off) to release the assistive force at the moment when S4 exceeded a threshold value (heel strike). The current of the soft actuators changed in accord with the variation of the applied DC voltage. These results demonstrated

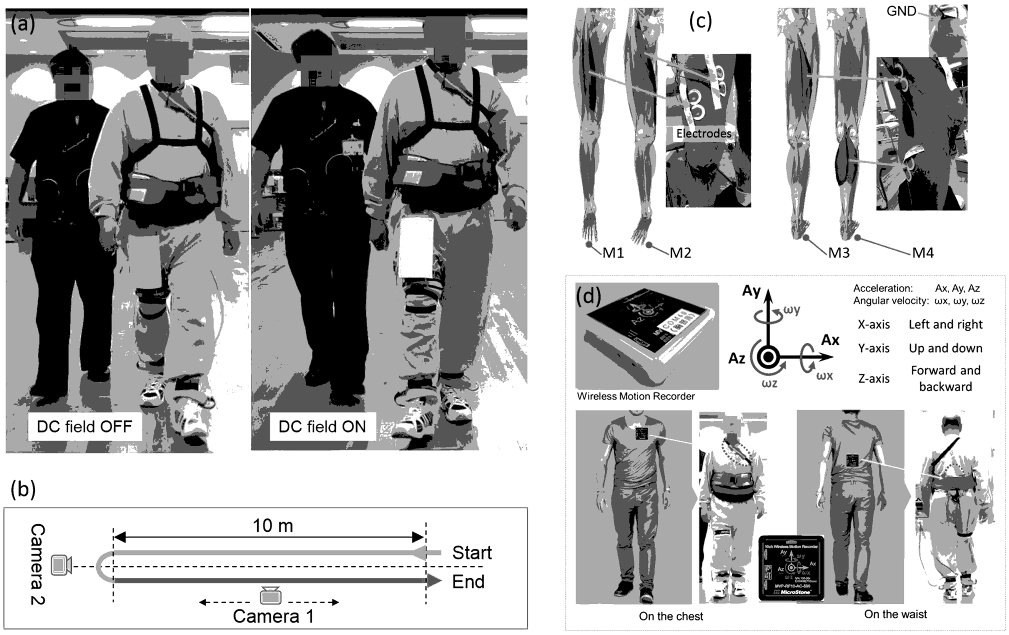

Figure 9. Overview of the walking assistance experiment. (a) Participant with the assist wear, accompanied by a physician during walking. (b) Conditions of the walking task. (c) EMG measurement of the femoris muscle (M1), sartorius muscle (M2), hamstring muscle (M3) and gastrocnemius muscle (M4). (d) Acceleration and angular velocity measurement of the chest and waist.

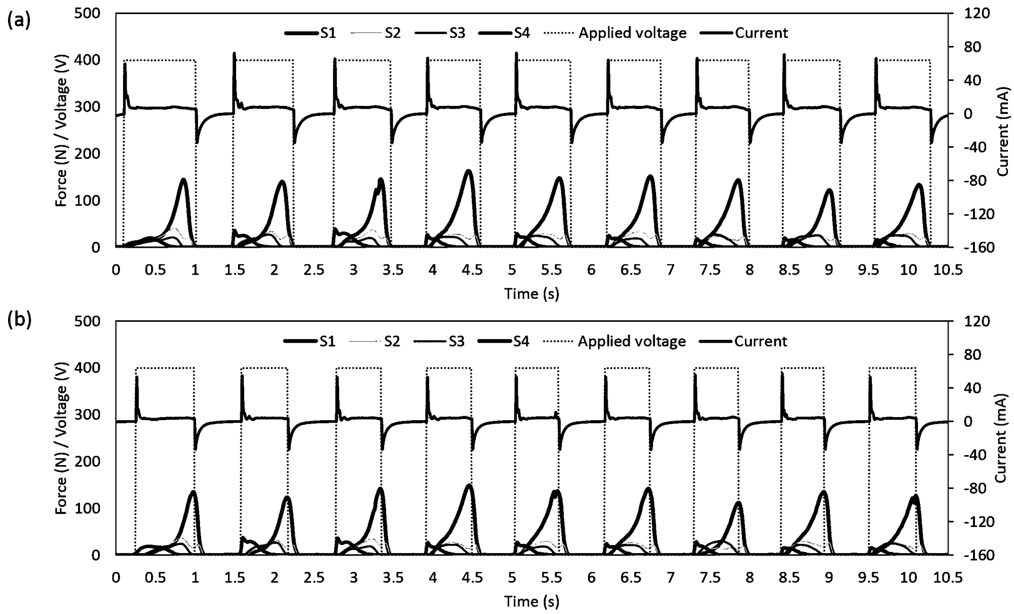

Figure 10. Results of walking motion detection and output voltage control in a 10 m one-way walking task. (a) Results of control strategy A. (b) Results of control strategy B.

that the walking motion detection and actuation components functioned effectively during the experiment.

4.2. Walking speed and step length

Figure 11 shows the average walking speed and step length in the experiment. Average walking speed increased by approximately 10% with the support of assist wear using control strategy B, with a significant difference of 5% by t-test. No significant difference was found for control strategy A (see figure 11(a)). This may have been because control strategy B provided an earlier assistance force before the leg left the ground in the pre-swing phase, which may have led to a longer supporting phase and a greater support force affecting the walking pace and step length. In addition, we found that the step length of the affected side increased by approximately 28 mm with the help of the assist wear using control strategy B, exhibiting a significant difference of 5% by t-test. No significant difference was observed on the healthy side (see figure 11(b)).

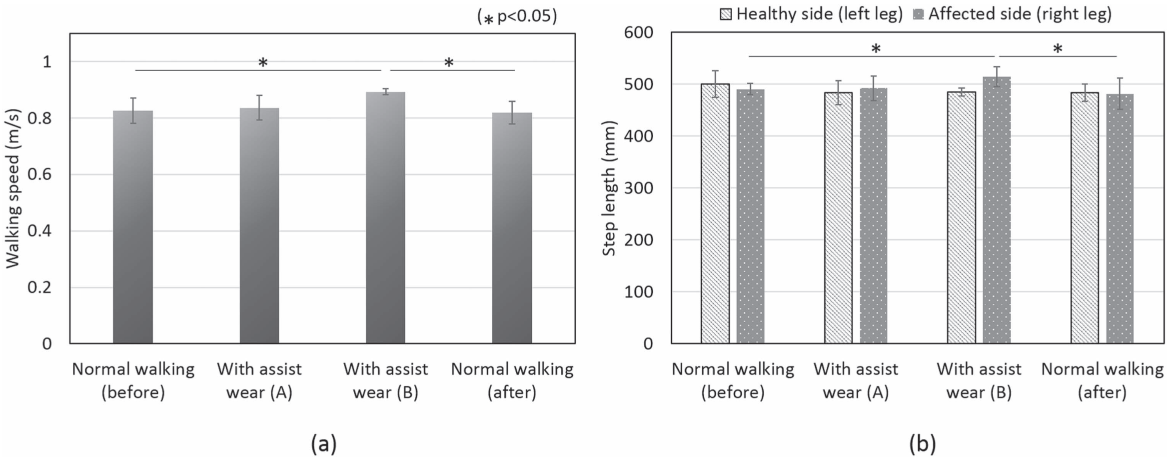

Figure 11. Average walking speed and step length of the participant with and without the assist wear. (a) Average walking speed. (b) Average step length.

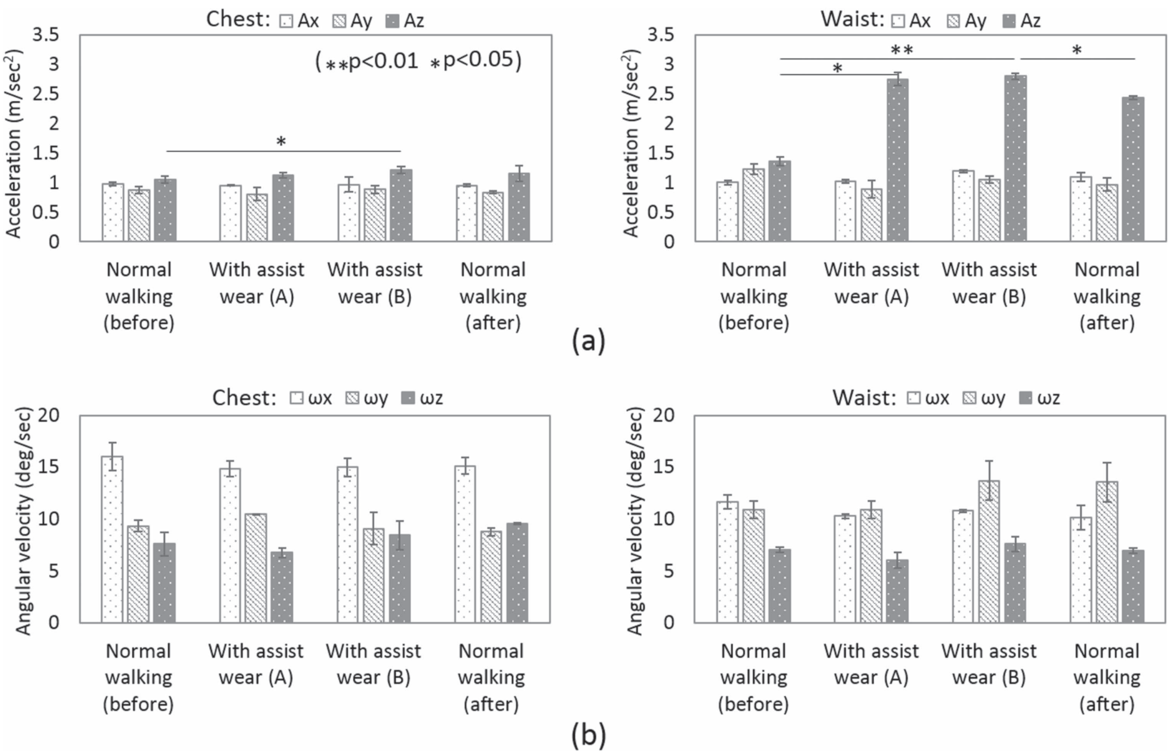

Figure 12. Acceleration and angular velocity of the upper body during walking. (a) Acceleration of the chest and waist. (b) Angular velocity of the chest and waist.

4.3. Acceleration and angular velocity

Figures 12(a) and (b) show the average results of the acceleration and the angular velocity of the upper body during walking, respectively. The acceleration in the forward and backward direction A_z was greater than that in the other directions, and increased with the assist wear using control strategy B both on the chest and waist, and control strategy A on the waist, with a significant difference of more than 5% by t-test (see figure 12(a)). This may indicate that with the support of the assist wear, the subject walked faster than normal walking speed without the assist wear, which is consistent with the walking speed results. No other significant differences were found in the left-right and up-down directions, possibly indicating that there was less negative influence of the assist wear on balance in the A_x and A_y directions. Regarding angular velocity (see figure 12(b)), no significant difference was found for the angular velocity in all directions. The angular velocity in the ω_x direction was greater than that in the other directions on the chest, possibly due to the body swing for a forward-backward rotation movement in straight line walking. On the waist, the angular velocity in the ω_x and ω_y directions was greater than in the direction of ω_z, and there was a tendency toward an increase when the assist wear was used (B) in the left-right rotation (ω_y) direction, possibly because of the assistance of the affected leg. These results

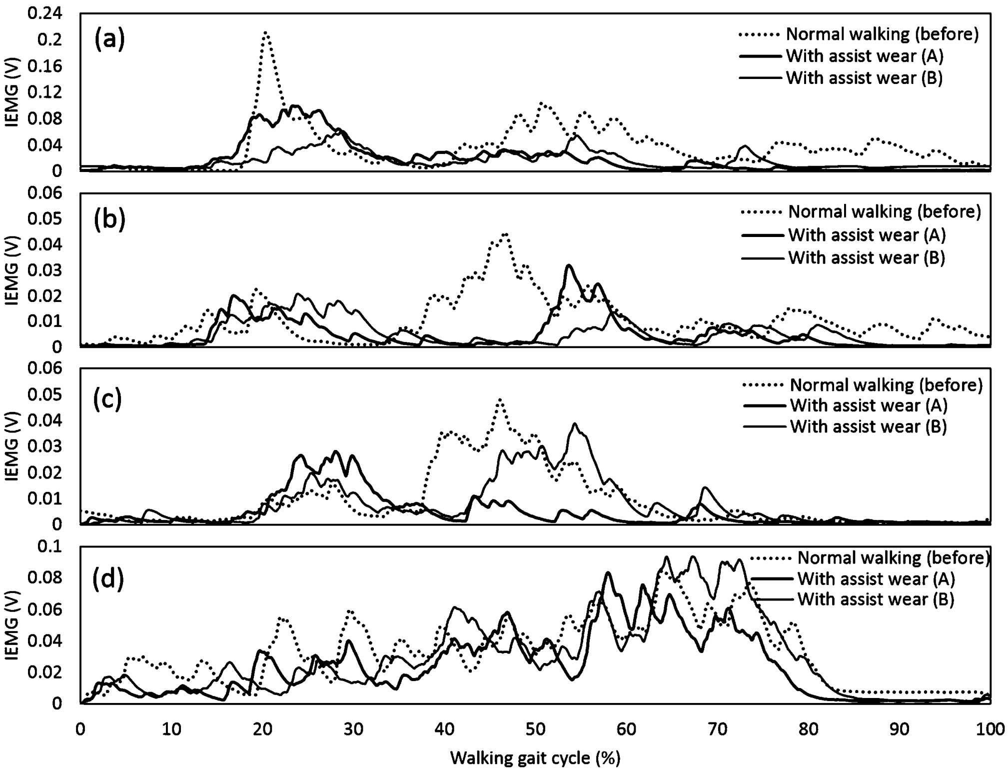

Figure 13. Integrated electromyogram (IEMG) variation during a walking gait cycle. (a) IEMG of the rectus femoris muscle. (b) IEMG of the sartorius muscle. (c) IEMG of the hamstring. (d) IEMG of the gastrocnemius.

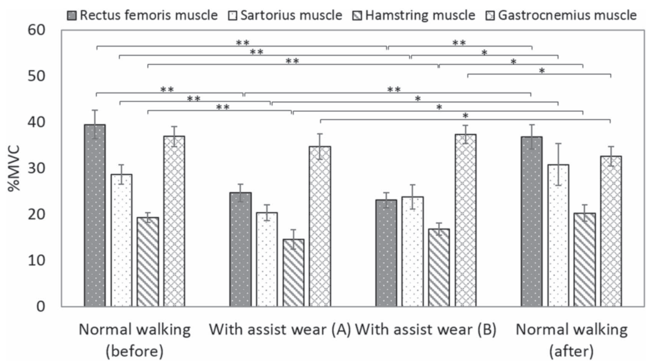

Figure 14. %MVC of the four types of muscles during walking with and without the PVC gel soft actuator-based assist wear.

may indicate that when wearing the PVC gel soft actuator-based assist wear, the wearer may be able to move faster while maintaining balance of the body during walking, enabling more efficient walking.

4.4. EMG variation

From the muscular activity variation shown in figure 13, it can be seen that the integrated electromyograms (IEMGs) of the rectus femoris muscle, the sartorius muscle and the hamstring muscle decreased in different degree during 40 to 60% of the gait cycle, with the assistance of the assist wear for both control strategies A and B. In addition, we found no significant change in the IEMG of the gastrocnemius muscle. We calculated the %MVC variation of each muscle using the average IEMG during walking and the maximal voluntary contraction IEMG measured before the experiment. We observed a maximum decrease of the %MVC of approximately 17% for the rectus femoris muscle, approximately 11% for the Sartorius, and approximately 5% for the hamstring, with a significant difference over 5% by t-test. However, the %MVC for the gastrocnemius muscle increased by approximately 5% compared with normal walking after the assistance experiment for both control strategies A and B (see figure 14). This was likely to be caused by the reduced amount of required muscular activity of the gastrocnemius with lower speed in normal walking after the assistance experiment. These EMG results indicate that the assist wear is effective for reducing the burden on the muscles of the lower limbs during walking.

4.5. Impression evaluation

Finally, the participant's subjective impressions of walking assistance with the PVC gel soft actuator-based assist wear were evaluated with a questionnaire. Each item was evaluated by the participant with a five-point scale from −2 to 2. Table 4 shows the results of the questionnaire after each walking task in the experiment. The participant reported that they walked freely with control strategy A, and felt assisted with control strategy B. The participant felt that the whole assist wear system was lightweight. These questionnaire results are consistent with the physical and physiological results described above.

In summary, the results indicate that the proposed PVC gel soft actuator-based assist wear is compact, lightweight and

Table 4. Impression evaluation questionnaire results.

Items	Score	
	Strategy A	Strategy B
Do you think you walked freely with the assist wear	1	0
Do you think you have been assisted (or walking easier) by the assist wear	−2	1
Do you feel you would be going to lose a balance during walking with the assist wear	−2	1
Do you think the PVC gel actuator based assist wear (including the power and controller) is light weight	2	

Table 5. A comparison between current actuator and improved actuator. When the PET-based electrode has almost the same mesh size and thickness with the current SUS-based electrode, by decreasing approximately 40% of the current thickness of the PVC gel membrane, we are able to decrease the total weight by about 50% and decrease the total thickness by about 40% but almost with the same displacement. Therefore, by adjusting the material and structure of the anode electrodes and the thickness of PVC gel membrane, we can improve and optimize the performance of PVC gel soft actuators.

	Current actuator	Improved actuator
Anode material	SUS 304	PET (with surface copper plating)
Anode mesh size/Thickness	#100/0.18 mm	#105/0.18 mm
PVC gel membrane thickness	200 μm	125 μm
Number of stacked layers	10 layers	10 layers
Applied voltage	400 V	400 V
Displacement/Contraction strain	0.58 mm/10%	0.57 mm/13%
Output force (when output displacement is about 0.11 mm)	12 N	18 N
Total weight	5 g	2.5 g

flexible, and does not impede the natural movement of the wearer during walking. The device enables the wearer to walk more easily and efficiently, and reduces the burden of the leg muscles during walking. The assistance effect can be adjusted by controlling the on-off time of the actuators. The current findings suggest the feasibility of the proposed assist wear for use in daily life.

5. Conclusions

Our long-term goal is to develop a lightweight, portable and compliant wearable walking assist wear device, suitable for use in daily life by older people. PVC gel soft actuator-based systems have substantial potential for supporting body motion with stable movement, quiet operation, considerable output force, low power consumption and a long cycle life. In addition, the shape profile design enables the device to be worn under clothing. In this preliminary study, we demonstrated that the system can enable natural walking without imposing constraints, as well as positively affecting walking speed and muscular activity.

However, for widespread practical application, a number of challenges require further research and development, such reducing the mass of the device and improving the output of PVC gel actuators, integrating multiple sensors, improving the human-machine interface and optimizing the control algorithm to provide more accurate and effective assistance for more joints and motions. Experiments with a greater number and range of subjects are required to confirm the performance of the assist wear system. To improve the current PVC gel soft actuator, we conducted other preliminary experiments and confirmed that by changing the current stainless mesh electrodes to conductive fiber mesh electrodes (e.g., resin fiber mesh with surface copper plating) together with decreasing approximately 40% of the current thickness of the PVC gel membrane, we were able to decrease the weight by more than 50%, and increase the performance (strain and stress) by more than 30% compared with the current actuator (see table 5). This may enable an assist wear device that is considerably lighter, reducing the burden for long-term use and providing greater assistance. Although we have focused on walking assistance for older people so far, there are a range of other potential applications for different types of motion assistance and rehabilitation. We speculate that soft actuator-based wearable soft robots may have widespread applications in the near future.

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