Development of Gait Training System Powered by Pneumatic Actuator like Human Musculoskeletal System

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Abstract—**The purpose of this study was to develop a body weight support gait training system for stroke and spinal cord injury (SCI) patient. This system consists of an orthosis powered by pneumatic McKibben actuators and a piece of equipment of body weight support. The attachment of powered orthosis can be fit to individual subjects with different body size. This powered orthosis is driven by pneumatic McKibben actuators arranged as a pair of agonistic and antagonistic bi-articular muscle models and two pairs of agonistic and antagonistic mono-articular muscle models like the human musculoskeletal system. The body weight support equipment suspends the subject's body in a wire harness, with the body weight is supported continuously by a counterweight. The powered orthosis is attached to the body weight support equipment by a parallel linkage, and its movement of powered orthosis is limited at the sagittal plane. The weight of the powered orthosis is compensated by a parallel linkage with a gas-spring. In this paper, we report the detailed mechanics of this body weight support gait training system and the results of several experiments for evaluating the system.**

Keywords; Body Weight Support, Locomotor Training

I. INTRODUCTION

The effect of body weight support treadmill training for incomplete SCI patient has been reported in previous studies. Wernig, et al. [1] demonstrated that 25 out of 33 persons with incomplete SCI were able to walk independently after treadmill training with partial body weight support. However, in this training process, two therapists must manually move both the patient's legs. Therefore, for the therapist, this training process is physically hard to continue for a long period. Colombo, et al. [2] developed a driven gait orthosis (DGO) that can be used on patients with varying degrees of paresis or spasticity for up to 30 min. Dietz, et al. [3] used this DGO on patients with incomplete SCI and suggested that the afferent input from lower limb and hip joints movement are important for the activation of central pattern generator for locomotion training in SCI patients.

We have already started to develop a gait training system powered by pneumatic McKibben actuators [4]. The advantages of this system are shown as follows: (1) McKibben actuators are arranged as two pairs of agonistic and antagonistic mono-articular muscle models and one pair of agonistic and antagonistic bi-articular muscle models like the human musculoskeletal system. This arrangement makes it possible to support the activity of paralyzed muscle using the corresponding part of McKibben actuator. Because an electric motor drives only for one joint, it is difficult to provide the support based on the degree of paralysis in each muscle. The McKibben actuator is very lightweight and can generate a large tensile force because it is pneumatic. In addition, it is possible to use McKibben actuator in water. We have already developed a gait training orthosis for use in water and observed that this orthosis is very effective at reducing enhanced hip extensor muscle activities. This is because this orthosis is meant for use in water and, consequently, the muscles of a patient undergoing training need to exert propulsive forces against water resistance [5, 6]. Therefore, the final goal of this study was to develop a body weight support gait training system for use in both on land and underwater environment. In this paper, we report the detailed mechanics of this body weight support gait training system and some of the results of several experiments for the evaluation of the land version.

II. SYSTEM DESIGN

A. Design of Body Weight Support Gail Training System

Fig.1 shows a schematic diagram of the body weight support gait training system. This system consists of a powered orthosis, treadmill, and body weight support equipment. The powered orthosis is fixed to the body weight support equipment by a parallel linkage. The weight of the powered orthosis is compensated by a gas spring, and its movement is limited at the sagittal plane. A counterweight is used to adjust the loading of the body weight support system in 1 kg increments. A subject wears a body harness and their vertical position is adjusted by a winch. The walking speed of the treadmill can be controlled manually and/or by a computer. The treadmill and pneumatic equipment are controlled by xPCTarget of MATLAB /Simulink (MathWorks Co.).
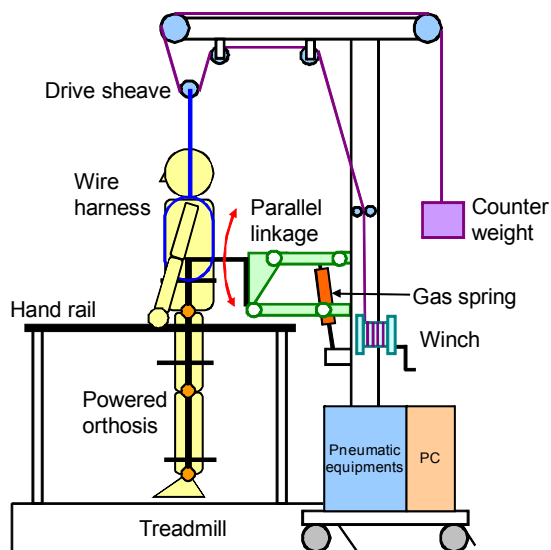
Fig. 1 Schematic diagram of body weight support gait training system. Counter weight can be set at maximum 100kg. Wire arrangement utilizes drive sheave and winch. Stroke of drive sheave is about 1500mm. Treadmill speed is from 0.1 to 6.0 km/h. Powered orthosis is suitable for 150 to 190 cm tall subject.

Fig.2 shows a picture of the developed body weight support gait training system. In the body weight support equipment, the counterweight follows the up-and-down trunk movement of a subject during walking. The wire arrangement uses a drive sheave, and it is possible to support the double the force of the counter weight. In addition, the drive sheave works as a wire tensioner. This mechanism can achieve the smooth movement of the counter weight during walking.

B. Pneumatic McKibben Actuator

The powered orthosis is driven by pneumatic McKibben actuators. These McKibben actuators are made in our laboratory (Fig.3) and have an effect similar to an actual human muscle contraction. Each expands to 1.0 [inch] across when pressurized. They generate a tensional force of approximately 800 [N] at 0.5 [MPa] without a load and have a 25% modulus of contraction from the normal length. They seem to be safe actuators because there is no potential for short circuits if used in water. In addition, because a pneumatic drive has no problem with oil leaks, air pressure is considered safer than hydraulic pressure.

C. Powered Orthosis

The powered orthosis for one limb utilizes two aluminum frames that can be adjusted for the thigh and leg lengths of each subject. A potentiometer is attached to the points where the axis passes through the hip and knee joints of the orthosis. Pneumatic McKibben actuators are attached as two pairs of agonistic and antagonistic mono-articular muscle models and

Fig. 2 Body weight support gait training system

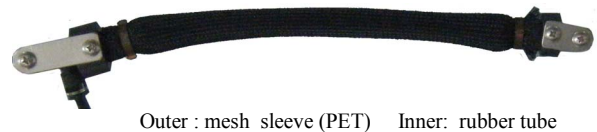Outer : mesh sleeve (PET)   Inner: rubber tube

Fig. 3 Pneumatic McKibben actuator. This is a linear-motion-type actuator. It has a 25% modulus of contraction from its normal length without a load.

one pair of agonistic and antagonistic bi-articular muscle models. The attachment positions of these McKibben actuators can also be changed to fit the body size of the subject.

D. Control System

Fig.4 shows a system diagram of the control and pneumatic equipment. The pneumatic control devices include a proportional directional control valve (MPYE5, FESTO Co.), electro-pneumatic regulator (ITV2030, SMC Co.), and pressure sensor (PSE540, SMC Co.). The air pressure is regulated from 0.8 [MPa] to under 0.5 [MPa] by a mechanical regulator (IR3020, SMC Co.). The proportional directional control valve controls the air supply to the two pneumatic McKibben actuators used in each agonistic and antagonistic mono-articular muscle model. The pressure of each actuator is measured by the pressure sensor. The control system is programmed using MATLAB/Simulink and is connected to the devices using xPCTarget with Simulink.

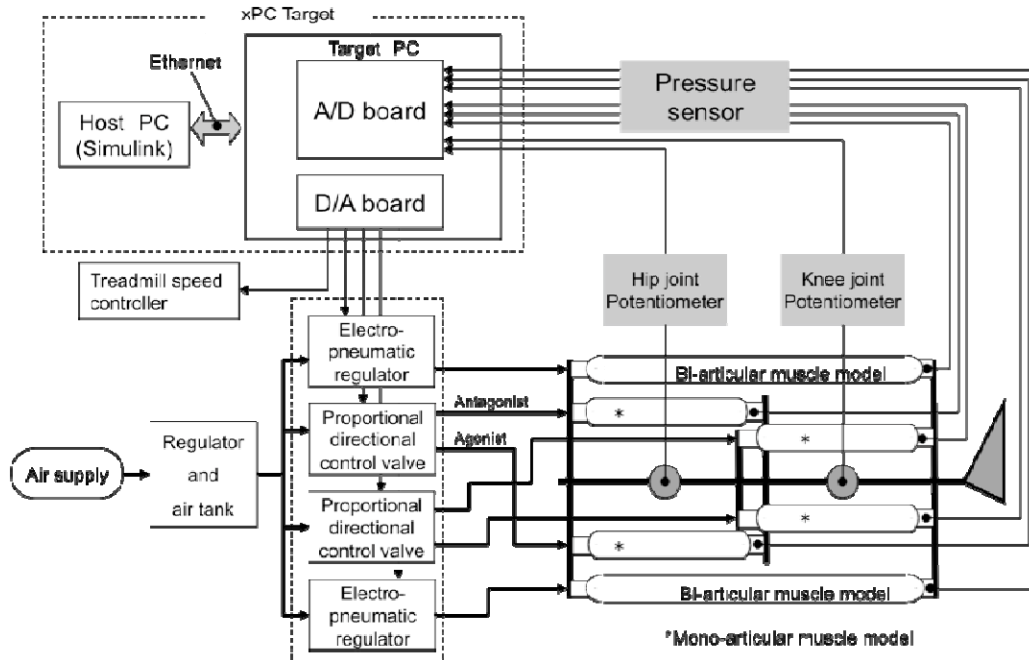The proportional directional control valve has two output

Fig. 4 System diagram of control and pneumatic equipments.

ports: port 2 and port 4. Output port 2 opens within a set range of 1 to 4.6 volts, and output port 4 opens within a set range of 5.6 to 10 volts. The reference point is approximately 5.2 volts. The reference point of the spool valve of the proportional directional control valve is located experimentally by varying the setpoint voltage of the proportional directional control valve from 4.5 to 5.6 volts in steps of 0.05 volts. However, even if the spool valve is located at the center position (reference point), the supplied air is not stopped completely; each actuator is always pressurized. Thus, the actuators of the mono-articular muscle model are always co-contracted. Therefore, the motions of the hip and knee joints of orthosis maintain stability by using this co-contraction of the actuators of the mono-articular muscle model.

One actuator of the bi-articular muscle model is controlled by one electro-pneumatic regulator adapted to the physiological timing of bi-articular muscle activation. The bi-articular muscle model is used to generate instantaneous force [7]. The actuators of the bi-articular muscle model generate the largest range of motion, as well as the strongest force and muscle moment.

Fig.5 shows a block diagram of the pressure feedback control system for the actuators of the mono-articular muscle model. The inner pressure difference between the agonist and antagonist actuators is calculated using Simulink. The calculated values are fed back to the PID controller. The input pressure data of the actuators of the mono-articular muscle model are used to calibrate the relationship between the pressure data and kinematic data of the human natural gait [8].
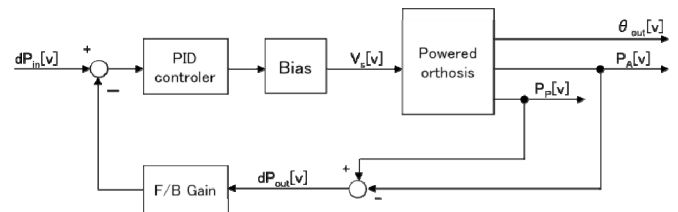

Fig.5. Control diagram for position and pressure
dP_{in} : Input data of inner pressure difference, dP_{out} : Inner pressure difference ($P_A - P_P$), V_s : Control voltage of MPYE valve, θ_{out} : Joint angle of powered orthosis

III. RESULTS AND DISCUSSION

A. Preliminary walking experiment

We attempted a preliminary experiment with powered orthosis walking. Fig.6 shows the kinematics data for a subject using the powered orthosis to walk with 100% of his body weight unloaded. The subject was instructed to rest completely, with no muscle activation. One gait cycle of the orthosis was 5 [s]. The maximum air pressure supplied was 0.5 [MPa]. The subject was a healthy male: Age, 21; Height, 170 [cm]; Weight, 61 [kg].

The broken line represents the input data for the human natural gait angle [8]. The thin and thick lines are the hip and knee joint angle data for just the mono-articular muscle model (thin line) and when applying both the mono- and bi-articular

muscle models (thick line) measured with the potentiometer of the orthosis. When the orthosis with the subject was driven using just the mono-articular muscle model, the range of motion for the hip joint was especially decreased as compared to that of the human natural gait angle pattern. This indicated that the flexion and extension forces of the hip joint were not satisfactory. In the case of applying both the mono- and bi-articular muscle models, it was not as decreased compared to using just the mono-articular muscle model. It seemed that adding the bi-articular muscle model increased the range of motion by the high muscle moment, especially the hip joint moment.

In this study, we tried to increase the stiffness of both the hip and knee joints by the co-contraction of agonistic and antagonistic mono-articular muscle models and to compensate for the lack of muscle moment by applying an agonistic and antagonistic bi-articular muscle model. On this point, the results of the preliminary experiment almost achieved the aim of the study.

However, the timing of the angle changes for both the hip and knee angles delayed the input data of the natural gait angle pattern. It seemed that this delay was caused by the mechanical property of the pneumatic actuator. It is necessary to improve the control system and/or to reform the structure of the pneumatic McKibben actuator.

Fig.6. Kinematics of powered orthosis.

IV. CONCLUSION AND FUTURE WORKS

We developed a body weight support gait training system powered by pneumatic McKibben actuators like a human musculoskeletal system. In our previous study, the subject's body size was restricted because the powered orthosis was based on general long leg braces with special fittings for individual subjects made by a prosthetist and orthotist. We developed a new powered orthosis that can be adjusted for individual subjects, along with body weight support equipment. This system showed good behavior in the case of a 100% unloading condition. However, the timing of the movement of the powered orthosis was not satisfactory compared to the natural human gait pattern. Future work will involve the following:

a) Determining the attachment position and natural length for the pneumatic McKibben actuator,
b) Establishing a precise position feedback control system for this powered orthosis, c) Developing a device for ankle foot support, d) Evaluating the system in several experiments, including the measurement of EMG, etc., e) Applying this system to underwater locomotor training.
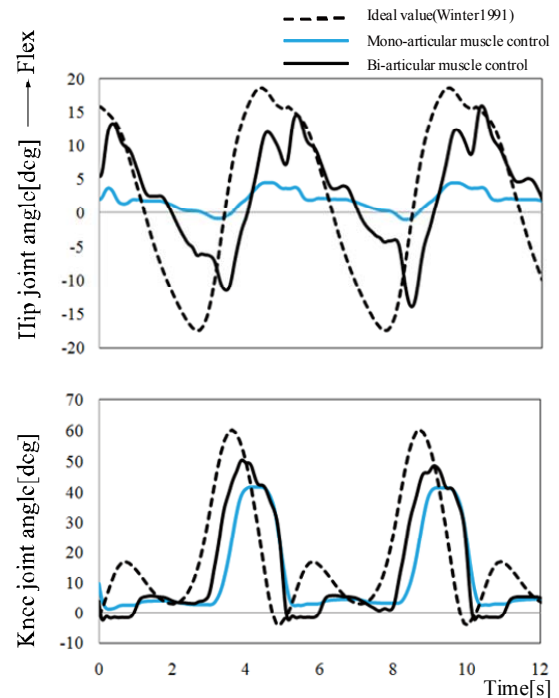
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