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Standing-Up Motion Support for Paraplegic Patient with Robot Suit HAL

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***Abstract*— This paper proposes a standing-up motion support system for complete paraplegic patients who cannot stand up by himself/herself due to a spinal cord injury. The standing-up motion is the first step for us to move somewhere inside and outside in our daily life. Therefore, the standing-up motion support is indispensable for patients to promote his/her independent life. The proposed support system using an exoskeletal assistive system "Robot Suit HAL" supports the wearer's weight during his/her standing-up motion so that he/she can stand up without any physical efforts. Besides, the support system controls the patient's posture for his/her stability to avoid falling down during the standing-up motion. The system also estimates his/her intention to stand up based on a preliminary motion of his/her upper body. The patient therefore starts the standing-up without any operations, just by bending his/her upper body forward as the preliminary motion. First, the system performance with respect to the weight-bearing and the balance control was confirmed through the experiment, by supporting a mannequin's standing-up. Then, the proposed system including the intention estimation algorithm was provided for a complete paraplegic patient in order to verify the performance of the total system. In consequence, we confirmed that the proposed system safely supported his standing-up according to his intention.**

(a) HAL-5 (Type-B) (b) HAL-5 (Type-C) (c) HAL-5 LB (Type-C)

Fig. 1. Robot suits "HAL-5". The HALs have been developed to assist physical capabilities of a wearer. This study uses the HAL-5 LB (Type C).

I. INTRODUCTION

There are a lot of people who have some troubles on their leg due to spinal cord injury (SCI), cerebrovascular accident (CVA) and so on. They may be forced to spend a wheelchair or bedridden life and they need care from others. Moreover, they tend to develop secondary troubles such as orthostatic hypotension, osteoporosis and bedsores, if they are forced to a bedridden life for a long term. These functional disease and physical damages limit his/her activity of daily living (ADL), and inflict a mental affliction. Therefore, these disorders are likely to lower his/her quality of life (QOL). For these reasons, the disorders of the functions in the lower limb movement make the patients fall into an adverse cycle. In order to enhance their ADL and to improve their QOL, it is important that they actively use their own legs in daily life. Particularly, the standing-up motion is extremely important, as the first step to make the patients promote their independent life. In the field of nursing care, doctors and physical therapists tend to emphasize the sit-to-stand training and the standing training as rehabilitation for the paraplegic patients who require prolonged bed rest. The training has several advantages e.g., expansion of the range of motion (ROM), activation of the circulatory system and the respiratory system, alleviation of spasticity and prevention of scoliosis. Especially in the training of standing-up motion using long leg braces, the advantageous effect is a remarkably higher bone mineral density (BMD) at the proximal femur [1]. Daily training also prevents orthostatic hypotension, osteoporosis and bedsores as side effects. The standing-up motion of patients however needs to be supported by doctors, physical therapists and own family as matters now stand. Consequently, a burden is imposed not only on the patients but also on caregivers.

On the basis of importance of the standing-up motion, various support systems that assist patients' standing-up have been therefore researched [2]–[6]. Most of these assistive devices support patient weight at any time such that a patient legs do not contribute to the motion. However it is better for an elderly person or a complete paraplegic patient to use their leg to support their weight due to the various reasons explained above. Noritsugu et.al., proposed a wearable power assist device using pneumatic rubber artificial muscles [7]. The assistive device shares patient weight with his/her legs and decreases knee joint torques of the wearer when his/her knee joints are bending. However, it does not contribute to his/her posture control due to lack of degrees of freedom. In addition to the weight-bearing, the posture control is also indispensable to stand up by himself/herself. We have developed an exoskeletal assistive system "Robot Suit HAL (Hybrid Assistive Limb)" shown in Fig. 1 in order to support the wearer's daily activities [8]–[10]. The "HAL-5 LB (Type-C)" used in this study supports the functional motions of lower limbs with the multiple joints simultaneously, using power units attached on hip, knee, and ankle joints. This paper proposes a balance control algorithm using these multiple degrees of freedom of the HAL during standing-up motion. In addition, a useful interface to directly convey

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the wearer's intention with regard to start or stop of the target motion to the assistive device such as a brain-computer interface is also desired, that is one of challenging issues. Bioelectrical signals such as surface EMG are one of the information to infer the wearer's intention related to his/her motion. Regrettably, the proper bioelectrical signals cannot be observed from patients such as a completely spinal cord injury patient. This paper therefore proposes an algorithm to estimate his/her intention related to start the standing-up based on a preliminary motion of his/her upper body. This preliminary motion can normally be observed in the initial phase of a standing-up motion of a person. The patient therefore starts standing-up, just by bending his/her upper body forward.

The purpose of this study is to realize a standing-up motion support system for complete paraplegic patients by using the HAL. The system starts supporting the wearer's weight according to his/her intention estimated by the proposed algorithm so that he/she can stand up intuitively. In addition, in order to stand up safely, we introduce a balance control algorithm to keep the wearer's stability based on his/her zero moment point (ZMP) [11]. To achieve the purpose mentioned above, this paper takes the following approaches:

(i) To confirm the effectiveness of the proposed balance control algorithm and the weight-bearing algorithm by a fundamental experiment in which the HAL makes a mannequin stand up from a sitting posture.
(ii) To adjust parameters of the control algorithm such as feedback gains, thresholds of the intention estimation and so on, by a preliminary experiment in which the system is applied to a healthy person who simulates the patient conditions.
(iii) To verify the performance of the proposed system including the intention estimation algorithm by a clinical trial in which a complete paraplegic patient stands up receiving a weight-bearing and a balance control.

II. Standing-up motion support system

In this section, we explain the "HAL-5 LB (Type-C)" used in this experiment, and an intention estimation that is one of the proposed algorithm.

A. Robot Suit HAL

The "HAL-5 LB (Type-C)" consists of power units, exoskeletal frames, sensors and a controller. Exoskeletal frames are fixed to the wearer's legs with molded fastening equipments. Potentiometers are attached to each joint to measure the relative angles. A triaxial accelerometer located in a control box measures the absolute angles of the wearer's trunk. The HAL can calculate the wearer's ZMP precisely by using the floor reaction force (FRF) sensors. These sensors utilizing semiconductor type pressure sensors are installed in the shoes. The weight of the HAL and the wearer is measured by the pressure of inner bags embedded into the plantar part of the shoes. A computer and batteries are attached on the wearer's waist, and motor drivers and other electrical circuits are allocated on each power unit. Power units are directly attached on each joint of the HAL. The actuators torque is transmitted from the HAL to the wearer's limbs through the mold fastening equipments.

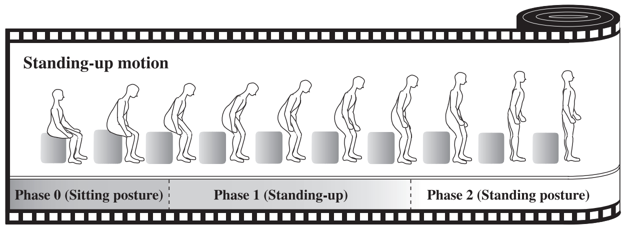


Fig. 2. Three phases in the standing-up motion. In "sitting posture phase," a wearer is sitting on a chair. The phase shifts to the following phase "standing-up" when a wearer's intention to stand up is observed. The next phase "standing posture" starts when a wearer has an upright posture.

B. Definition of three phases at the target motion

We have proposed a "Phase Sequence" concept that combines some motion elements as short-term phases into a sequence of a functional motion in order to achieve a target motion [10]. For example, a biped walk is divided into two phases from the viewpoint of contact conditions; single support phase and double support phase. A biped walk is realized by combining the single support phase and the double support phase. In the same way, we divide the standing-up motion including immediately before and after the motion into three phases; sitting posture phase, standing-up phase, and standing posture phase as shown in Fig. 2. These phases are defined as follows:

1) The trunk inclines forward during sitting posture (Phase 0).
2) The buttocks leave the seating surface (Phase 1).
3) The knee and the hip joints extend in order to stand upright (Phase 1).
4) Maintaining an upright posture (Phase 2).

Transfer conditions from phase 0 to phase 1 are the inclination angle of the patient's upper body and the position of the ZMP. The transfer condition from phase 1 to phase 2 is the knee joint angle of the patient.

C. Intention estimation algorithm

The ultimate interface would be to directly convey the wearer's intention with regard to the target motion to the assistive device such as a brain-computer interface. Detecting the bioelectrical signals such as surface EMG is one of the ways to infer the wearer's intention related to his/her motion in the form of electrical potential. Unfortunately the proper bioelectrical signals cannot be observed from patients such as a completely spinal cord injury patient. In another way, we propose an estimation algorithm to infer the intention of the patient from the preliminary motion which is observed intuitively before the target motion. As a result, the system starts the standing-up motion support synchronizing with his/her motion.

A person inclines his/her upper body forward before his/her standing-up in order to transfer a position of center of gravity (COG) of his body forward enough to support his weight on his legs stably. The ground projection of the COG corresponds to a center of pressure (CoP) in a static posture. In addition, the CoP also corresponds to the ZMP [12]. For this reason, the patient's intention to stand up is detected by the body inclination and the ZMP transfer. The patient therefore starts the standing-up without any operations, just by bending his/her upper body forward as the preliminary motion.

The system estimates that a patient intends to start the standing-up, when the following inequalities are satisfied:

$$\theta_{hip} > \theta_{thre\_hip} \quad \text{and} \tag{1}$$

$$x_{zmp} > x_{thre\_zmp}, \tag{2}$$

where θ_{thre_hip} and x_{thre_zmp} are thresholds that are adjusted in the preliminary experiment. The phase 0 shifts to the phase 1 once the intention to stand up is estimated. Next, the phase 1 shifts to the phase 2 when the below inequality is satisfied:

$$\theta_{knee} < \theta_{thre\_knee}, \tag{3}$$

where θ_{thre_knee} is threshold to shift to the phase 2. When the condition shown inequality (3) is satisfied, the HAL starts the standing posture support.

III. Controller design

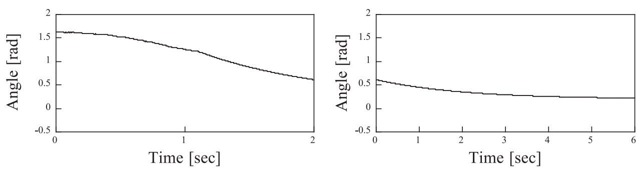
In this section, we explain the controller design for the standing-up motion support system. The control strategy consists of two algorithms, that is, a balance control and a weight-bearing control. The following sub-sections explain the details of these algorithms.

A. Control strategy

The proposed support system provides two algorithms as follows to support the patient's standing-up motion.

- Balance control algorithm based on the wearer's ZMP.
- Gravity compensation algorithm for the weight-bearing.

The torque of each joint of the HAL is calculated by considering two required algorithms; the balance control algorithm and the gravity compensation algorithm. The angle of the ankle joints have much influence on the position of the ZMP [13]. The range of the torque applied to the ankle joints is however limited because the feet start rotating around the tips of the toes or the heels. The hip joints in addition to the ankle joints are also used to control the position of the ZMP as the balance control. On the other hand, the knee joints are used to lift up the COG of the patient. A height transition of the COG as well as the position of the ZMP is very important for the wearer to feel comfortable during standing-up motion. The knee joints are used to control the height of the COG, because they directly contribute to the height of the COG. Additionally, it is necessary to provide a suitable assist for the wearers' physical characteristic, so as not to give the wearer an uncomfortable feeling. Therefore, the reference trajectory of the height is extracted from a healthy person's standing-up motion. The extracted motion pattern shown in Fig. 3 is applied to phase 1 and phase 2. The time period of the trajectory and the amplitude of the trajectory can be adjusted to the wearer's physical characteristic, the degree of disorder and the wearer's preference.



(a) (b)

Fig. 3. Reference patterns of knee joint angle. (a) in the standing-up phase, (b) in the standing posture phase.

The ankle joint torques are sum of the PD control of the ZMP measured by the FRF sensors (Figure 4 (a)) and the gravity compensation as shown in eq. (4). The ZMP control and the gravity compensation are explained in the following subsections. The hip joint torques are the sum of the PD control of the ZMP and the gravity compensation as shown in eq. (5). The reference angle of the hip joint θ_{ref_hip} is calculated from the kinetic model (Figure 4 (b)) such that the current ZMP goes toward the reference ZMP position. The detail is explained in the next subsection. The knee joint torques are the sum of the PD control using the reference pattern based on a healthy person's knee trajectory and the gravity compensation as shown in eq. (6).

$$\tau_{ankle} = \boldsymbol{K_{Pa}}(x_{ref\_zmp} - x_{zmp}) - \boldsymbol{K_{Da}}\dot{x}_{zmp} + \tau'_{ankle}, \tag{4}$$

$$\tau_{hip} = \boldsymbol{K_{Ph}}(\theta_{ref\_hip} - \theta_{hip}) - \boldsymbol{K_{Dh}}\dot{\theta}_{hip} + \tau'_{hip} \quad \text{and} \tag{5}$$

$$\tau_{knee} = \boldsymbol{K_{Pk}}(\theta_{ref\_knee} - \theta_{knee}) - \boldsymbol{K_{Dk}}\dot{\theta}_{knee} + \tau'_{knee}, \tag{6}$$

where τ_{ankle}, τ_{hip}, τ_{knee}, $\tau'_{ankle,}$, $\tau'_{hip,}$, $\tau'_{knee,}$, x_{zmp}, θ_{hip}, θ_{knee}, $\dot{x}_{zmp}$, $\dot{\theta}_{hip}$, $\dot{\theta}_{knee}$, x_{ref_zmp}, θ_{ref_hip} and θ_{ref_knee} are column matrixes. These variables have two elements that corresponded to each leg, respectively. Feedback gains $\boldsymbol{K_{Pa}}$, $\boldsymbol{K_{Da}}$, $\boldsymbol{K_{Ph}}$, $\boldsymbol{K_{Dh}}$, $\boldsymbol{K_{Pk}}$ and $\boldsymbol{K_{Dk}}$ are diagonal matrixes where feedback gains for each leg are diagonal elements. They are given as follows:

$$\begin{aligned}
\boldsymbol{K_{Ph}} &= \begin{bmatrix} K_{Ph_R} & 0 \\ 0 & K_{Ph_L} \end{bmatrix}, \boldsymbol{K_{Dh}} = \begin{bmatrix} K_{Dh_R} & 0 \\ 0 & K_{Dh_L} \end{bmatrix}, \\
\boldsymbol{K_{Pk}} &= \begin{bmatrix} K_{Pk_R} & 0 \\ 0 & K_{Pk_L} \end{bmatrix}, \boldsymbol{K_{Dk}} = \begin{bmatrix} K_{Dk_R} & 0 \\ 0 & K_{Dk_L} \end{bmatrix}, \\
\boldsymbol{K_{Pa}} &= \begin{bmatrix} K_{Pa_R} & 0 \\ 0 & K_{Pa_L} \end{bmatrix}, \boldsymbol{K_{Da}} = \begin{bmatrix} K_{Da_R} & 0 \\ 0 & K_{Da_L} \end{bmatrix},
\end{aligned} \tag{7}$$

where subscripts Ph_R, Ph_L, Pk_R, Pk_L, Pa_R and Pa_L mean the proportional gain of the right hip joint, the left hip joint, the right knee joint, the left knee joint, the right ankle joint and the left ankle joint, respectively. Dh_R, Dh_L, Dk_R, Dk_L, Da_R and Da_L mean the derivative gain of the right hip joint, the left hip joint, the right knee joint, the left knee joint, the right ankle joint and the left ankle joint, respectively.

B. Balance control algorithm based on wearer's ZMP

Three FRF sensors are installed in the toe part, the ball part and the heel part of the sole one by one. The ZMP in back and forth direction is calculated by:

$$x_{zmp} = \frac{\Sigma f_i x_i}{\Sigma f_i}, \tag{8}$$

where x_{zmp} is the ZMP in the sagittal plane. f_i are sensor outputs of each part and x_i are sensor positions on the x-axis (Figure 4 (a)). According to biomechanical analysis, the ZMP of a healthy person is located around the ankle joint axis from the viewpoint of the ankle joint torque. In this study, the reference of ZMP position x_{ref_zmp} is located however 30∼50 [mm] in front of the ankle joint axis from the viewpoint of stability [14]. As a result, some dorsal flexion moment is necessary in the ankle joints. The stability however becomes the maximum in back and forward directions. The ankle joints are directly controlled based on the error of the ZMP position as shown in the first term on the right side of eq. (4).

The hip joints are also controlled based on the ZMP. The reference angle of the hip joints, however, is calculated from the kinetic model as shown in Fig. 4 (b), in order to respond to the offset caused by the difference between individual physical conditions. In addition, the hip joints are controlled based on the angular error of the hip joint as shown in the first term on the right side of eq. (5). The ZMP position in the sagittal plane x'_{zmp} is calculated based on a direct kinematics method expressed by:

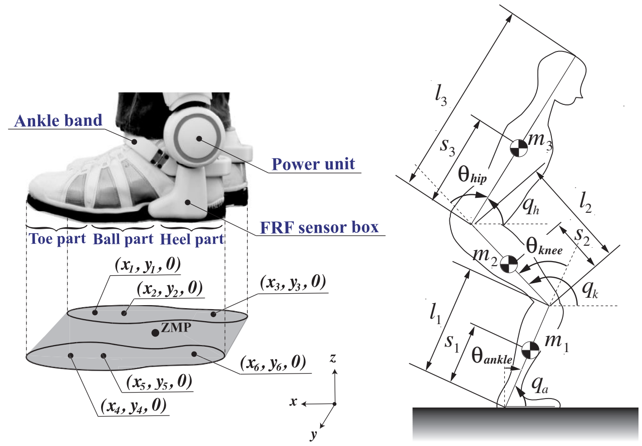
$$\begin{aligned} x'_{zmp} = \frac{1}{m_1+m_2+m_3} \Big\{ &(m_1 s_1 + m_2 l_1 + m_3 l_1)\cos(\frac{\pi}{2} - \theta_{ankle}) \\ &-(m_2 s_2 + m_3 l_2)\cos(\frac{\pi}{2} - \theta_{ankle} + \theta_{knee}) \\ &+m_3 s_3 \cos(\frac{\pi}{2} - \theta_{ankle} + \theta_{knee} - \theta_{hip}) \Big\}, \end{aligned} \tag{9}$$

where m_i is the mass of link i, s_i is the position of mass i and l_i is the link length, respectively.

The inverse kinematics of the hip joint angle θ_{hip} are solved by using eq. (9). The reference hip joint angle θ_{ref_hip} is decided uniquely, if the reference of the ZMP position x'_{ref_zmp} is substituted for x'_{zmp}. Therefore, θ_{hip} can be expressed as θ_{ref_hip} shown by:

$$\begin{aligned} \theta_{ref\_hip} = &\frac{\pi}{2} - \theta_{ankle} + \theta_{knee} \\ &-\cos^{-1}\Bigg[\frac{1}{m_3 s_3}\Big\{(m_1+m_2+m_3)x'_{ref\_zmp} \\ &+(m_2 s_2 + m_3 l_2)\cos(\frac{\pi}{2} + \theta_{ankle} - \theta_{knee}) \\ &-(m_1 s_1 + m_2 l_1 + m_3 l_1)\cos(\frac{\pi}{2} - \theta_{ankle})\Big\}\Bigg], \end{aligned} \tag{10}$$

where θ is the relative angle of each link. The reference hip joint angle θ_{ref_hip} shown in (10) is updated at each control cycle based on the current other joint angles. Therefore, disturbance control can be performed.



(a) (b)

Fig. 4. (a) Three floor reaction force (FRF) sensors embedded in the toe part, the ball part and the heel part of the sole. (b) Definition of system parameters and variables. The flexion direction of each joint angle is set to be positive and each joint angle becomes 0 [rad] in an upright posture.

C. Gravity compensation algorithm for weight-bearing

The high gains of the PD control in eqs. (4)-(6) are necessary in order to lower the error from the reference angles, if a constant large force such as a gravity affects the system joints. The gravity compensation of the patient mass and the system mass enables to be fixed lower gains of the PD control so that the stiffness of the system joints can be lower. That contributes to support the patient's motion with flexibility. The gravity compensation torque of each joint is calculated by:

$$\tau'_{hip} = -m_3 s_3 g cos q_h, \tag{11}$$

$$\tau'_{knee} = -\{(m_3 l_2 + m_2 s_2) g cos q_k + m_3 s_3 g cos q_h\}, \tag{12}$$

$$\begin{aligned} \tau'_{ankle} = -\{&(m_1 s_1 + m_2 l_1) g cos q_a + \\ &(m_2 s_2 + m_3 l_2) g cos q_k + m_3 s_3 g cos q_h\}, \end{aligned} \tag{13}$$

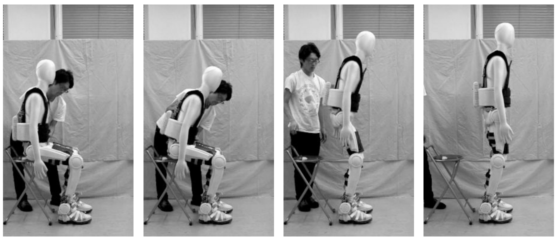
where q is the absolute angle of each link.

IV. Experiments & Results

The proposed algorithm for the standing-up motion support was verified through three types of experiments. At first, the effectiveness of the proposed balance control algorithm and the gravity compensation algorithm were confirmed through a fundamental experiment in which the HAL makes a mannequin stand up from a sitting posture. Next, parameters of the control algorithm were adjusted through a preliminary experiment in which the support system is applied to a healthy person who simulates the patient's condition. Finally the performance of the proposed algorithm including the intention estimation algorithm was verified through a clinical trial where a complete paraplegic patient stands up receiving the weight support and the balance control.

A. Fundamental experiment

1) Experimental settings: In this fundamental experiment, the HAL makes a mannequin stand up from a sitting posture.



(a) Phase 0 (b) Phase 1 (c) Phase 1 (d) Phase 2

Fig. 5. Sequential photographs during the standing-up motion support by the HAL. The HAL starts standing-up at phase 1, when the start conditions are satisfied. The joint torques of the HAL are conveyed to the mannequin through the belts that fasten the mannequin at several points. During standing-up motion, the hip joints and ankle joints are controlled by keeping the ZMP of the total system at the reference point. The knee joints are controlled to follow the reference trajectory extracted from a healthy person's motion. The HAL supports standing posture in phase 2 after finishing standing-up.

The mannequin is used in order to simulate the condition of a complete paraplegic patient. In addition, it does not have any actuator and remains passive such as a complete paraplegic patient without contracture. The performances of the balance control algorithm and the gravity compensation algorithm are investigated without any disturbances such as a wearer.

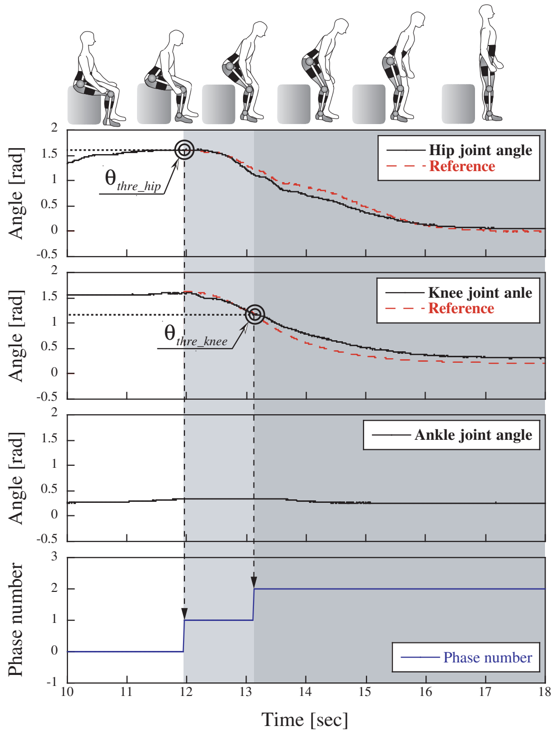
Normally, the HAL starts supporting the mannequin's standing-up motion when the conditions shown in inequalities (1) and (2) of the intention estimation algorithm are satisfied. The mannequin cannot take action using the upper body with a cane or a walker to satisfy the conditions by itself. A man therefore pushes the upper body of the mannequin in the horizontal direction to bend the hip joints. Once the conditions are satisfied, the man releases his hand from the mannequin to stop pushing.

2) Result: Figure 5 shows the sequential photographs of the mannequin's standing-up motion supported by the HAL. Figure 6 (a) shows joint angles, references and phase transitions. From the results of the joint angles in these figures, the hip and knee joints followed the reference angles. The ZMP and the range of the reference ZMP during each phase are shown in Fig. 6 (b). In phase 1, the reference trajectory x_{ref_zmp} initially starts at 9.8 [cm] and terminates at 5.0 [cm]. Besides, the reference trajectory x_{ref_zmp} moves from 5.0 [cm] to 2.0 [cm] in phase 2. The result of our experiment clearly showed that the mannequin's posture could be controlled with the two algorithms. In consequence, we confirmed that the HAL supported standing-up properly, without other support-equipments.

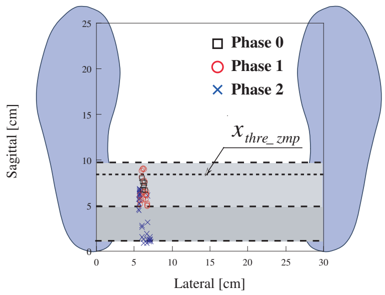
B. Preliminary experiment

1) Experimental settings: As a preliminary step towards a clinical trial with the paraplegic patient, the proposed system is applied to the healthy person who has similar physical parameters to the patient. Table I shows the parameter settings of the patient. These parameters were defined on the basis of each body segment. Additionally, the patient in this study has completely impaired motor and sensory functions of the lower limbs. Therefore, the healthy person completely relaxes the legs to simulate the patient's lower limbs function. In order to acquire appropriate feedback parameters, the standing-up motion support is repeatedly performed until each joint follows the reference patterns generated by the proposed method. The PD gains of the standing-up motion support in eq. (7) are designed in this experiment. Feed back gains are given by K_{Ph_*}, K_{Pk_*}, K_{Pa_R}, K_{Dh_*}, K_{Dk_*} and K_{Da_R}, respectively.

2) Result: Figure 7 shows joint angles, references, ZMP, phase transitions and torques of the power units during the standing-up motion support. Torques in each joint were calculated using eqs. (4), (5) and (6). From the results of these figures, his hip, knee and ankle joints follow the reference. That means the HAL successfully supported his standing-up based on the proposed method. Consequently,



(a) Joint angles in each phase



(b) ZMP transition in each phase

Fig. 6. Result of the fundamental experiment using a mannequin with free joints. These data show the angle of each joint, distribution of the ZMP, and the transition of each phase. These results show that the HAL stably controlled the mannequin's motion from sitting posture to standing-up posture.

TABLE I

PARAMETER SETTINGS OF THE PATIENT.

Mass [kg]		Length [m]		Length [m]	
m_1	6.62	s_1	0.25	l_1	0.45
m_2	6.64	s_2	0.2	l_2	0.35
m_3	38.00	s_3	0.45	l_3	0.80

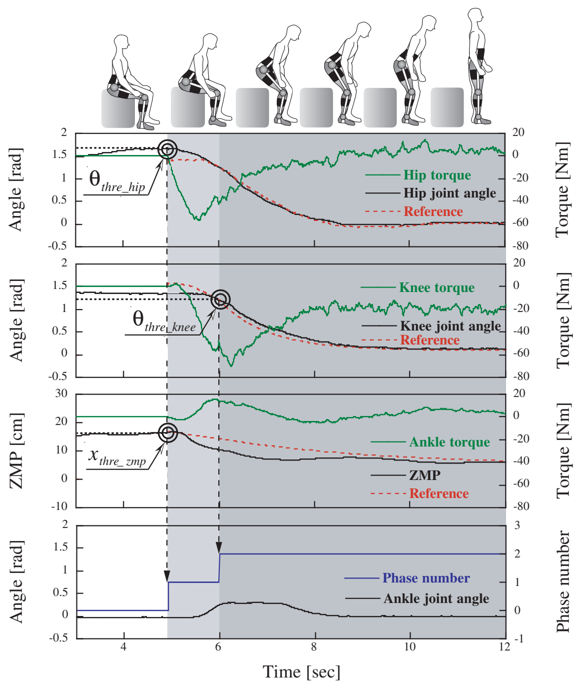


Fig. 7. Result of the preliminary experiment. These graphs show the angle and the torque of each joint, the ZMP, and the transition of each phase. The torque of each joint was calculated using eqs. (4), (5) and (6). His hip, knee, and ankle joints almost follow the reference trajectories in phases 1 and 2.

the gains in phase 1 and phase 2 were determined as $K_{Ph_*} = 130.0$, $K_{Dh_*} = 2.0$, $K_{Pk_*} = 150.0$, $K_{Dk_*} = 2.8$, $K_{Pa_*} = 30.0$ and $K_{Da_*} = 1.2$. As a result, we could start the clinical trial for the complete paraplegic patient by using these parameter settings.

C. Clinical trial

1) Experimental settings: Figure 8 shows the experimental setting in this clinical trial. The HAL supports the paraplegic patient. The participant is a 66-years-old male, 160 centimeters tall and his weight is 68 kilograms. He is a spinal cord injury patient having almost all of his motor and sensory functions on the lower body impaired. He is diagnosed with complete spinal cord injury because the tenth and eleventh thoracic vertebrae (T10 and T11) are damaged due to the vertebral fracture.

He has given informed consent before participating in this clinical trial. All procedures were approved by the "Institutional Review Board", and this clinical trial was conducted under the inspection of a medical doctor. The physical condition in his lower limbs on the day was ex-

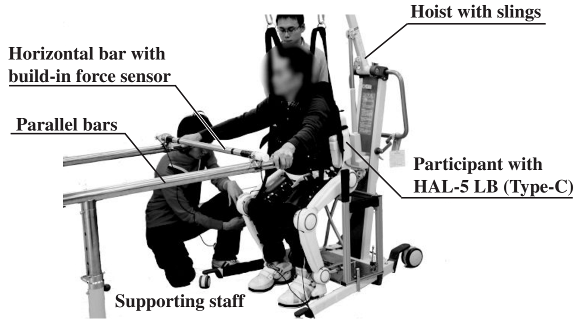


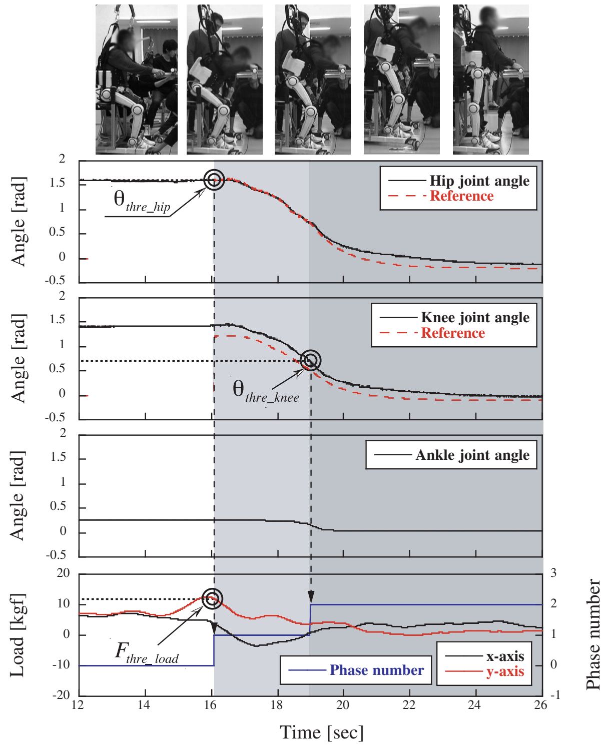
Fig. 8. Experimental setting in the clinical trial. A hoist is connected with the waist of the patient trough a sling slackly because the hoist is prepared to prevent the patient from falling down in the case of a system failure.

amined by the doctor. Furthermore, the velocity of reference knee joint angle in phase 1 and reference angles in phase 2 were adjusted in advance so as to prevent muscle spasm from developing in the clinical trial. After the advance preparation mentioned above, the proposed system supported his standing-up motion. In consequence, muscle spasticity in his lower limbs which might restrict the use of the system was not observed during the clinical trial.

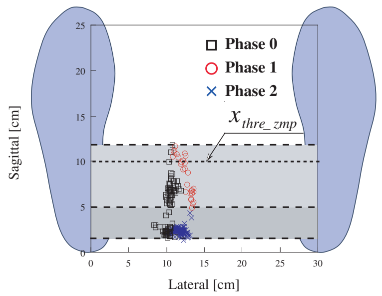
The waist sling installed on his torso is connected with a hoist in order to prevent an emergency fall during this trial. However, the belt that connects the hoist with the waist sling is normally loose case so as not to disturb the standing-up posture of the patient. Additionally, a horizontal bar with built-in force sensor is fixed to parallel bars 90 centimeters height in front of the participant so that he could incline his upper body easily by pulling the bar with both his arms. The force sensor measures the force applied to the bar by him. In addition to inequalities (1) and (2), the load applied to the bar is used for the intention estimation from phase 0 to phase 1 for safety. The chair is 45 centimeters height, and is fixed to the hoist so as not to move.

2) Result: Figure 9 (a) shows joint angles, reference angles and phase transitions during the standing-up motion support. The ZMP and the range of the reference ZMP during each phase are shown in Fig. 9 (b). In phase 1, the reference trajectory x_{ref_zmp} initially starts at 12.5 [cm] and terminates at 5.0 [cm]. Later, the reference trajectory x_{ref_zmp} moves from 5.0 [cm] to 2.5 [cm] in phase 2. The HAL estimated that he intends to stand up from a sitting posture when inequalities (1), (2) and the threshold of the bar in the y-axis were satisfied. After that, the HAL started the standing-up motion support (Phase 1). Next, when the inequality (3) was satisfied during phase 1, phase 2 started. From the results of the joint angles in these figures, hip and knee joints followed the reference trajectories. Moreover, the ZMP followed the reference ZMP. Consequently, the result of the ZMP measurement provide stability of his balance during standing-up motion support.

The load applied to the bar was used as an additional transition condition from phase 0 to phase 1 for safety. Furthermore the load in the vertical direction (y-axis) was



(a) Joint angles in each phase



(b) ZMP transition in each phase

Fig. 9. Result of the clinical trial. These data show the angle of each joint, the ZMP, the load on the bar and the transition of each phase. The condition shown in the inequality (1) and the threshold of the bar in the y-axis were satisfied (15.8 [sec]), however, the condition shown in the inequality (2) was not satisfied. For this reason, the proposed support system did not start the standing-up motion. When the inequality (2) was also satisfied (16.0 [sec]), the system estimated that the patient intended to start the standing-up. From these results, the proposed system safely achieved to support his standing-up motion according to his intention.

below 8 [kgf] in phase 1, and below 4 [kgf] in phase 2. This result shows that the standing-up motion support by the HAL was achieved without excessive force to his upper body.

V. Conclusions

In this paper, we proposed a standing-up motion support system for paraplegic patients by using the proposed algorithm. The effectiveness of the proposed balance control algorithm and the weight-bearing algorithm were confirmed through a fundamental experiment. The results of our experiment clearly showed that the HAL stably controlled the mannequin's motion from sitting posture to standing-up posture. In a preliminary experiment, parameters such as feedback gains and thresholds of the intention estimation were adjusted appropriately, in order to find the best parameter settings for the patient in advance. In the clinical trial for the complete spinal cord injury patient with sensory and motor paralysis, we verified the performance of the proposed system including the intention estimation algorithm. Consequently, the total system safely achieved to support his standing-up motion. Through these stages, we realized a standing-up motion support system for complete paraplegic patients by using the proposed algorithm.

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