The hip joints are also controlled based on the COP. The reference angle of the hip joints, however, is calculated from the kinetic model as shown in Fig. 5, in order to absorb the offset caused by the difference between individual physical parameters. 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 (8). The position of COP in the sagittal plane x'_{cop} is calculated based on a direct kinematics method expressed by:

$$\begin{aligned} x'_{\mathrm{cop}} = \frac{1}{m_1+m_2+m_3}\bigg\{&(m_1s_1+m_2l_1+m_3l_1)\cos\left(\frac{\pi}{2}-\theta_{\mathrm{ankle}}\right) \\ &-(m_2s_2+m_3l_2)\cos\left(\frac{\pi}{2}-\theta_{\mathrm{ankle}}+\theta_{\mathrm{knee}}\right) \\ &+m_3s_3\cos\left(\frac{\pi}{2}-\theta_{\mathrm{ankle}}+\theta_{\mathrm{knee}}-\theta_{\mathrm{hip}}\right)\bigg\}, \end{aligned} \tag{13}$$

where θ_{ankle} is the relative angle of ankle joint as shown in Fig. 5, 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 (13). The reference angle of the hip joint $\theta_{\mathrm{ref\_hip}}$ is obtained uniquely, if the reference position of the COP $x'_{\mathrm{ref\_cop}}$ is substituted for x'_{cop}. Therefore, $\theta_{\mathrm{ref\_hip}}$ is obtained by:

$$\begin{aligned} \theta_{\mathrm{ref\_hip}} = \frac{\pi}{2}-\theta_{\mathrm{ankle}}+\theta_{\mathrm{knee}}-\cos^{-1}\bigg[\frac{1}{m_3s_3}\bigg\{&(m_1+m_2+m_3)x'_{\mathrm{ref\_cop}} \\ &+(m_2s_2+m_3l_2)\cos\left(\frac{\pi}{2}+\theta_{\mathrm{ankle}}-\theta_{\mathrm{knee}}\right) \\ &-(m_1s_1+m_2l_1+m_3l_1)\cos\left(\frac{\pi}{2}-\theta_{\mathrm{ankle}}\right)\bigg\}\bigg]. \end{aligned} \tag{14}$$

The reference hip joint angle $\theta_{\mathrm{ref\_hip}}$ shown in (14) is updated at each control cycle based on the current other joint angles.

4.3. Gravity Compensation Algorithm for Weight Bearing

The high gains of the PD control in (7)–(9) are necessary in order to lower the errors from the reference angles if a constant large force such as gravity affects the system joints. The gravity compensation of the patient's weight and the system's mass enables us to fix lower gains of the PD control so that the stiffness of the system joints could be lower. That contributes to supporting the patient's motions with flexibility. The gravity compensation torque of each joint is calculated by:

$$\begin{aligned} \tau'_{\mathrm{ankle}} = -\bigg\{&(m_1s_1+m_2l_1)g\cos\left(\frac{\pi}{2}-\theta_{\mathrm{ankle}}\right) \\ &+(m_2s_2+m_3l_2)g\cos\left(\frac{\pi}{2}-\theta_{\mathrm{ankle}}+\theta_{\mathrm{knee}}\right) \\ &+m_3s_3g\cos\left(\frac{\pi}{2}-\theta_{\mathrm{ankle}}+\theta_{\mathrm{knee}}-\theta_{\mathrm{hip}}\right)\bigg\} \end{aligned} \tag{15}$$

$$\tau'_{\mathrm{hip}} = -m_3s_3g\cos\left(\frac{\pi}{2}-\theta_{\mathrm{ankle}}+\theta_{\mathrm{knee}}-\theta_{\mathrm{hip}}\right) \tag{16}$$

$$\begin{aligned} \tau'_{\mathrm{knee}} = -\bigg\{&(m_3l_2+m_2s_2)g\cos\left(\frac{\pi}{2}-\theta_{\mathrm{ankle}}+\theta_{\mathrm{knee}}\right) \\ &+m_3s_3g\cos\left(\frac{\pi}{2}-\theta_{\mathrm{ankle}}+\theta_{\mathrm{knee}}-\theta_{\mathrm{hip}}\right)\bigg\}, \end{aligned} \tag{17}$$

where g is the gravitational acceleration.

5. Experiments

The proposed algorithms for the sit-to-stand and stand-to-sit transfer support are verified through experiments that are separately executed in two steps from the viewpoint of safety. In the first step, parameters of the control algorithms are adjusted through a preliminary experiment because stiffness and damping factors of a human body could not be modeled precisely. A healthy person wearing the HAL simulates the conditions of a patient who has completely impaired motor and sensory functions of the lower limbs. He, however, keeps his body balance by himself, controlling his lower limbs in unexpected situations. The experiment for parameter settings could be conducted safely, thanks to the high adaptability of the healthy person. In the following step, the patient safely receives the motion support from the HAL without overshooting from the beginning of the clinical trial by using the adjusted parameters.

5.1. Preliminary Experiment

5.1.1. Experiment Settings

A healthy person who has similar physical parameters to the patient was adopted as a subject in the preliminary experiment. The HAL's weight of each link is measured in advance. However, the precise wearer's weight of each body segment is not measurable. In order to apply the gravity compensation algorithm to the patient with the HAL, the wearer's weights of each body segment are necessary as well as the HAL's weights of each link. The mean values of elderly Japanese men from statistics as shown in Table 1 are, therefore, used for the balance control algorithm and the gravity compensation algorithm. The values are previously calibrated by the

Table 1.
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	48.00	s_3	0.45	l_3	0.80

experiment so as not to give an uncomfortable feeling to the wearer. The healthy person completely relaxes his legs to simulate the lower limb functions of the patient who has completely impaired motor and sensory functions of the lower limbs. The proportional gains K_{Ph_*}, K_{Pk_*} and K_{Pa_R} and derivative gains K_{Dh_*}, K_{Dk_*} and K_{Da_R} used in the feedback control (10) are adjusted so that each joint could follow the reference trajectory without overshooting.

5.1.2. Results

The PD gains of each phase were adjusted in the preliminary experiment. Figure 7 shows the angles of the hip and knee joints, the COP trajectory, and those references through three phases during sit-to-stand transfer support. These graphs show that the angles of the hip and knee joints and the COP follow the references without overshooting when the feedback gains are set as $K_{Ph_*} = 100.0$, $K_{Dh_*} = 3.5$, $K_{Pk_*} = 130.0$, $K_{Dk_*} = 3.5$, $K_{Pa_*} = 18.0$ and $K_{Da_*} = 8.0$. Those gains are fixed in phase 1, phase 2 and phase 3.

5.2. Clinical Trial

5.2.1. Experimental Environment

In this paper, proposed algorithms are applied to a paraplegic patient. The participant is a 66-years-old male, 160 cm tall and his weight is 68 kg. He is diagnosed with complete SCI because the T10 and T11 thoracic vertebrae are damaged due to vertebral fracture. He can control his posture using parallel bars with both his arms so as to convey his intention related to the sit-to-stand and stand-to-sit transfers as shown in Fig. 8. Additionally, a horizontal bar is fixed to the parallel bars at a height of 90 cm in front of the patient so that he can grip the bar in an unexpected situation such as an emergency fall. The waist sling installed on his torso is connected to a hoist. However, the belt that connects the hoist with the waist sling is normally slack so as not to disturb sit-to-stand and stand-to-sit transfers of the patient. In addition, the height of the chair has the greatest influence on the motion support because the knee flexion moment is reduced by raising the height of a chair [28]. In this clinical trial, the height of the chair was set at the hight that corresponds to 100% (45 cm height) of the length of the patient's lower thigh. The chair is fixed to the hoist to make sure it does not move during the motion support.

The patient gave 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 of his lower limbs was examined by a medical doctor just before every trial. Furthermore, the maximum velocity of reference knee joints angles was adjusted in advance to prevent muscle spasm from developing during the clinical trial. After the preparations mentioned above, muscle spasticity in his lower limbs, which might restrict the use of the system, was not observed during the clinical trial.

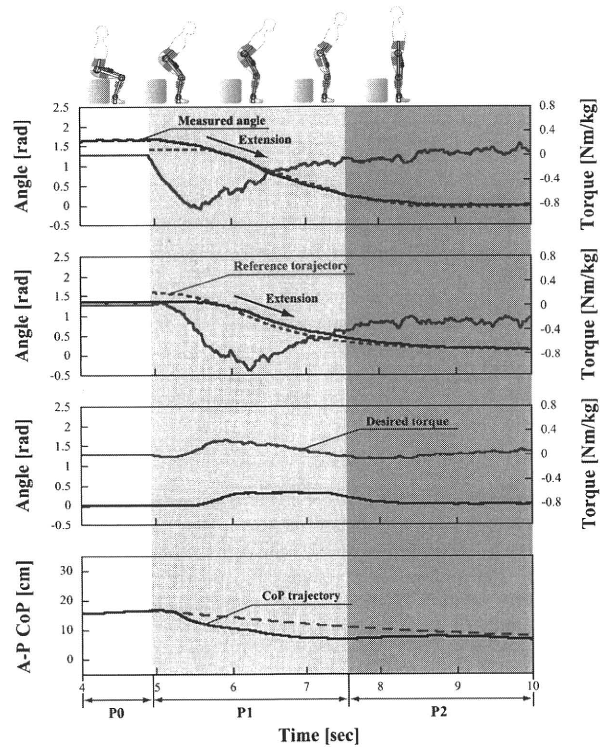

Figure 7. Result of the preliminary experiment. These graphs show the angle and torque of each joint and the COP calculated by the FRF sensors in the A–P direction through three phases. The desired torque of each joint is calculated using (7), (8) and (9). The torque of each joint is a normalized value based on the weight of the total system. P0 (white area), P1 (light gray area) and P2 (gray area) denote the phase of sitting, sit-to-stand transfer and standing, respectively. Dashed lines represent the reference trajectories. The ankle joint is controlled so that the current COP of the total system could follow the reference of the COP. On the other hand, the reference trajectory of the knee joints is calculated based on a healthy person's sit-to-stand transfer. The reference angle of the hip joint is calculated based on the angles of the ankle joint and the knee joint using (14).

5.2.2. Results

We verified the performance of the proposed algorithms including the intention estimation algorithm through a clinical trial with the complete SCI patient. Figure 9

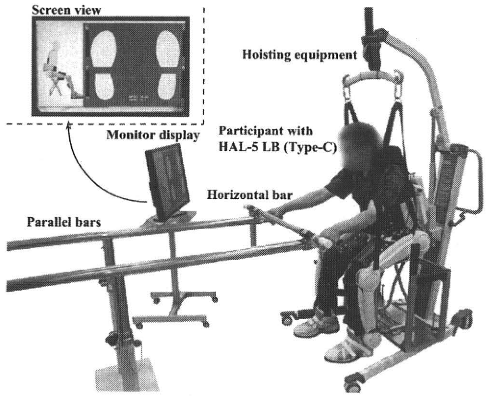

Figure 8. Experimental environment of the clinical trial. A hoist is connected to the waist of the patient through a slack sling so that the hoist could prevent the patient from falling down in the case of system failure. Joint angles of the wearer's lower limbs and the COP of the total system are displayed on a monitor display during sit-to-stand and stand-to-sit transfers.

shows the angles and torques of each joint, the FRFs and the COP through three phases during sit-to-stand transfer. The results in these graphs indicate that the hip and knee joints follow the reference trajectories. In addition, the COP also follows the reference trajectory. The COP in the anterior–posterior (A–P) and medial–lateral (M–L) directions during sit-to-stand transfer is shown in Fig. 10. After the buttocks left the seating surface (phase 1), the system is able to control the COP of the total system within 38.2–58.7% of the support polygon in the A–P direction from 3.8 to 6.8 s. After the sit-to-stand transfer was completed (phase 2), the system is able to control the COP within 31.8–38.2% of the support polygon in the A–P direction from 6.8 to 16.0 s. These results indicate that the COP control keeps the stability of his balance during sit-to-stand transfer support.

Figure 11 shows the COP trajectory and the phase transitions at the start of the stand-to-sit transfer. In order to start the stand-to-sit transfer phase during phase 2, the wearer shifts his COP backward by pushing his upper body with his arms slightly. The HAL estimates that he intends to sit down from a standing posture when inequality (5) is satisfied. Then, the HAL starts his stand-to-sit transfer support, synchronizing his intention. Figure 12 shows the angles and torques of each joint and the COP through three phases during stand-to-sit transfer support. The results in these graphs indicate that the hip and knee joints follow the reference trajectories. The COP also follows the reference trajectory. The COP in the A–P and M–L directions during stand-to-sit transfer is shown in Fig. 13. In phase 2, the system is able to control the COP within 21.3–35.6% of the support polygon in the

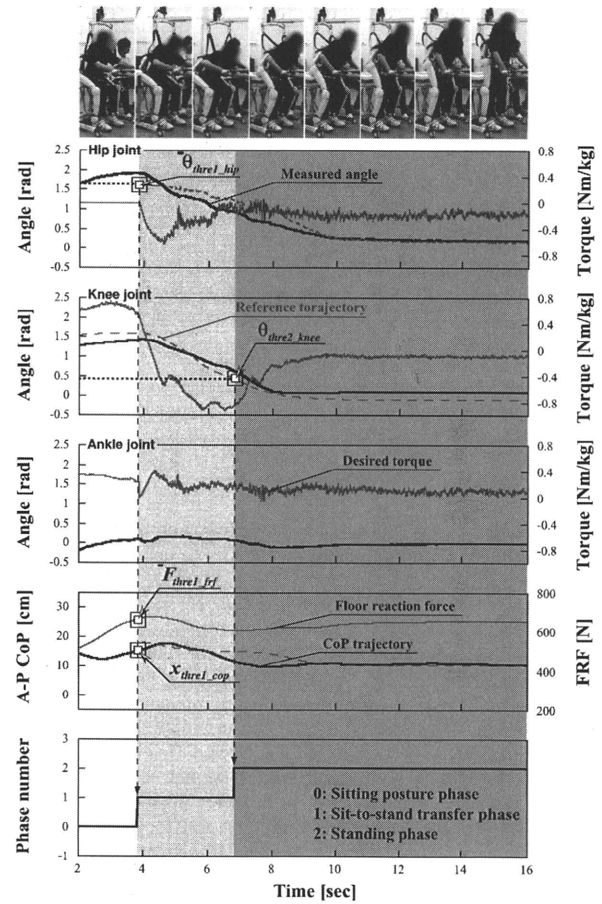

Figure 9. Result of the sit-to-stand transfer support in the clinical trial. These graphs show the angle and torque of each joint, the FRFs, and the COP calculated by the FRF sensors in the A–P direction through three phases. The desired torque of each joint is calculated using (7)–(9). The HAL estimated that the wearer intends to stand up from a sitting posture when inequalities (1)–(3) were satisfied at 3.8 s. At that time, the HAL started the sit-to-stand transfer phase (phase 1). Next, when inequality (4) was satisfied during phase 1, the standing phase (phase 2) started at 6.8 s.

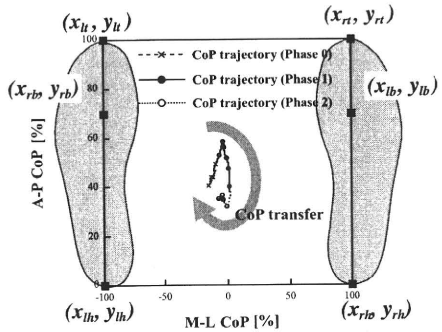

Figure 10. COP trajectory of the support polygon in the A–P and M–L directions during sit-to-stand transfer support.

A–P direction from 18.0 to 19.7 s. In phase 3, the system is able to control the COP within 17.7–37.9% of the support polygon in the A–P direction from 19.7 to 26.5 s. These results indicate that the COP control keeps the stability of his balance during stand-to-sit transfer support.

Figure 14a shows the mean position of the COP in the A–P direction from phase 1 to phase 2 when the wearer received the sit-to-stand transfer support from the HAL while looking at the monitor that shows the current COP and joint angles of the wearer's lower limbs. The proposed support system can control the COP at 39.7% (standard deviation: 9.71%) on average. As shown in Fig. 14b, the system controls the COP at 40.1% (standard deviation: 13.04%) on average when he received the sit-to-stand transfer support from the HAL without looking at the monitor. Figure 15a shows the mean position of the COP in the A–P direction from phase 2 to phase 3, when the wearer received the stand-to-sit transfer support from the HAL while looking at the monitor. The system can control the COP at 32.4% (standard deviation: 7.95%) on average. As shown in Fig. 15b, the system controls the COP at 25.3% (standard deviation: 10.31%) when he received the stand-to-sit transfer support from the HAL without looking at the monitor. The results mean that the patient who confirms his current COP depicted on the monitor could control his balance by himself during the transfers cooperating with the HAL so that stability and safety could be increased.

6. Discussion

Physical support during sit-to-stand and stand-to-sit transfers is important for the independent life of paraplegic patients. In particular, we focused on the motion support for complete paraplegic patients. Although an assistive system should start

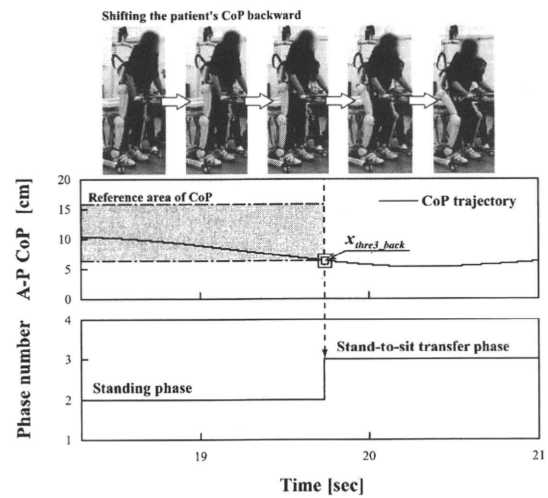

Figure 11. Detection of the wearer's intention to sit down. Sequential photographs show how the wearer pushes the parallel bars with his arms to convey his intention related to sit down to the HAL and that he starts the stand-to-sit transfer. In phase 2, the COP of the total system was controlled within the range of the reference COP indicated in the gray area. After that, he shifted the COP to the threshold x_{thre3_back} at 19.7 s in order to start the stand-to-sit transfer. Then, the HAL started supporting his stand-to-sit transfer, synchronizing his intention. The range of the reference COP in the A–P direction is set from 6.35 to 15.4 cm based on the result of the preliminary experiment empirically.

supporting the motions at the proper moment synchronizing the patient's intention to stand up and to sit down, it has not yet been solved. The purpose of this study was to realize the control method of complete paraplegic patients during sit-to-stand and stand-to-sit transfers through an intuitive interface.

To achieve this purpose, we proposed the algorithms to support the wearer's weight and body posture for stability, and to infer the intention based on a preliminary motion that is observed just before sit-to-stand and stand-to-sit transfers. The proposed algorithms embedded in the HAL were applied to a complete SCI patient who is able to shift his COG by using upper body functions including arms and hands. The physical condition of the wearer's lower limbs should be examined before its use so as to prevent any muscle spasm from developing.

In the clinical trial, we confirmed that the wearer could intuitively start the sit-to-stand and stand-to-sit transfer support based on a preliminary motion of his upper body instead of his bioelectrical signals. The HAL detected the wearer's intention to stand up from the preliminary motions of his body trunk inclining forward and

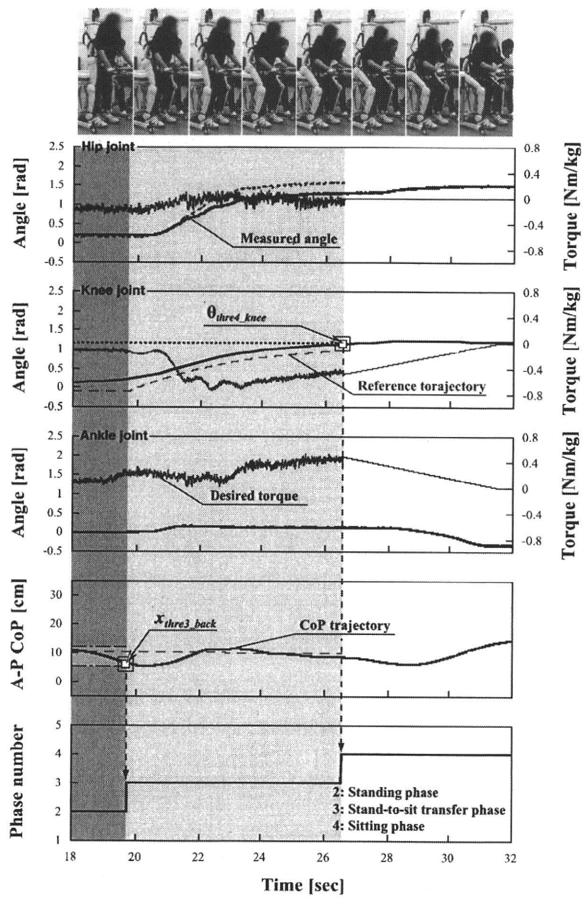

Figure 12. Result of the stand-to-sit transfer support in the clinical trial. These graphs show the angle and torque of each joint and the COP calculated by the FRF sensors in the A–P direction through three phases. The HAL estimated that the wearer intends to sit down when the conditions shown in inequality (5) was satisfied (19.7 s). At that time, the HAL started the stand-to-sit transfer phase (phase 3). Next, when the condition shown in inequality (6) was satisfied during phase 3, the phase shifted to the sitting phase (phase 4) at 26.5 s. In phase 4, the torque of each joint of the HAL is designed to decrease gradually so that the wearer could smoothly sit on the chair.

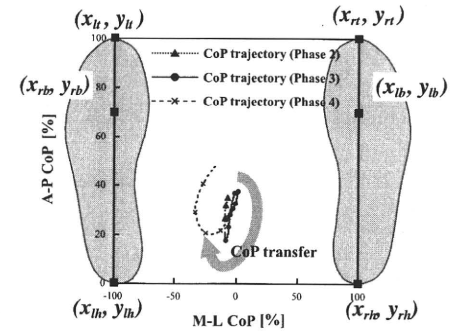

Figure 13. COP trajectory of the support polygon in the A–P and M–L directions during stand-to-sit transfer support.

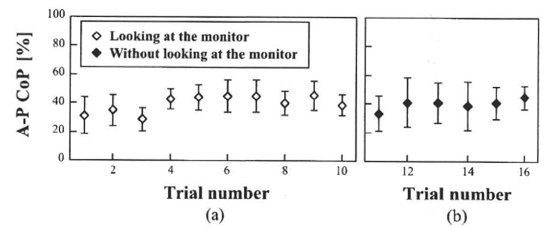

Figure 14. Mean position of the COP of the support polygon in the A–P direction during sit-to-stand transfer support: (a) while looking at the monitor and (b) without looking at the monitor.

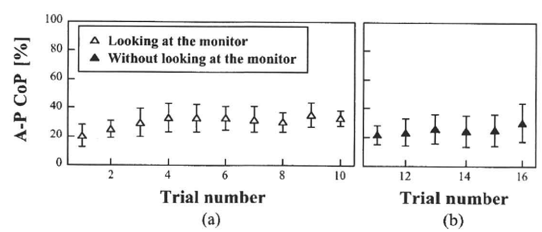

Figure 15. Mean position of the COP of the support polygon in the A–P direction during stand-to-sit transfer support: (a) while looking at the monitor and (b) without looking at the monitor.

the COP exceeding the threshold, and it autonomously supported the patient's sit-to-stand transfer. In addition, the HAL detected the wearer's intention to sit down from the preliminary motions that made the COP go backward out of the support polygon when the wearer pushed his upper body with his arms during the standing phase (phase 2). After that, the HAL autonomously supported the patient's stand-to-sit transfer. These results indicated that the proposed algorithms successfully estimated his intention to stand up and to sit down.

During sit-to-stand transfer, the proposed system controlled the COP from 17.6 to 55.6% of the support polygon in the A–P direction as shown in Fig. 14a. During stand-to-sit transfer, the proposed system controlled the COP from 12.7 to 43.7% of the support polygon in the A–P direction as shown in Fig. 15a. The current COP information depicted on the monitor in front of the patient contributes to the precision of COP control in both transfers from the viewpoint of the standard deviations of the COP as shown in Figs 14b and 15b.

The most significant factor of these improvements is that the patient could understand the current condition of his balance from the monitor instead of the sensory feedback from the lower limbs. We are developing an auditory feedback system for complete paraplegic patients that would have practical implications in daily life, because the visual feedback occupies a patient's vision for interaction with his environment.

7. Conclusions

We proposed an algorithms to support the wearer's weight and to control the wearer's body posture for stability during sit-to-stand and stand-to-sit transfers. In addition, we proposed an algorithm to estimate the wearer's intention to start these motions based on a preliminary motion of their upper body and posture condition. In a clinical trial with a complete SCI patient, we confirmed that the proposed algorithms support sit-to-stand and stand-to-sit transfers of the patient with the HAL safely and conveniently by keeping his stability and by reflecting his intentions. The results of this study show that we realized the control method of complete paraplegic patients during sit-to-stand and stand-to-sit transfers by using the proposed algorithms.

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PRACTICAL DESIGN OF FULL BODY EXOSKELETONS

Stretching the Limits of Weight and Power

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Abstract: The development of full body, energetically autonomous exoskeletons has been limited by the constraints of weight and available power. Because of this it has not been possible to create one that augments all DoF of its human wearer with enough power to assist, e.g., nurses and rescue workers. To achieve more usefulness despite the limitations, a practical design approach that considers the motions and needs of the wearer is an appropriate tool to reveal new opportunities. This approach was used to find solutions for a fully supported 3DoF exo-spine, supported shoulder girdle motion, and other challenges that have so far received little or no attention. No extra actuators are required, thus adding a minimum to weight and power. The improvements found using this practical approach suggest related fields like rehabilitation could profit as well.

1 INTRODUCTION

Recent levels of technology have enabled many researchers to make various exoskeletal devices: robots that surround (parts of) a human wearer in order to assist him in his movements. Applications range from rehabilitation to strengthening nurses and others in their work. Yet, the all-round applicability, and thus usefulness, of wearable, energetically autonomous (W-EA) exoskeletons, has so far been limited by the low amount of degrees of freedom (DoFs) and actuators achievable in such devices.

Realizing maximum usefulness requires augmentation of all human DoFs. This necessitates many heavy actuators and an accordingly large, heavy power supply for a useful running time of perhaps a few hours. The more useful and thus larger the device the more unlikely it is to fit in the settings of a hospital or home, and hence we are faced with a challenge to find the optimal balance.

Considering the needs of aging societies to take care of the older generations, this research focuses on exoskeletons for augmentation of nurses and other workers, and has a long term goal to also develop a version for physically challenged patients. It is based on the full body robot suit HAL-5 from which a lower body suit was derived for patients who have difficulty walking (Suzuki et al. 2005).

This paper will therefore review existing exoskeletons, show why and how we should shift our approach to achieving solutions, and finally, to show its appropriateness, propose solutions from a mechanical perspective that maximize the capabilities of these full body robot suits.

1.1 Existing Exoskeletons

Lower body exoskeletons have been discussed (Dollar & Herr 2008) and few challenges remain. As for the upper body part (from the hip to the hands) there are more DoFs and larger workspaces that

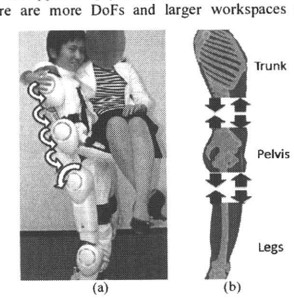


Figure 1. Hip and trunk interdependence during lifting. In (a) the HAL-5 suit supports the moments in the hips and back as indicated by the arrows. The interaction forces indicated in (b) show their interdependence.

ultimately compete with the constraints of weight and power. Hence, this is the focus for new solutions.

One W-EA full body exoskeleton is the nurse power suit (Yamamoto 2002). It uses a pneumatic actuator system to augment the muscles used for lifting patients. While it is focused on patient lifting, its workspace, however, is otherwise limited.

Another, the Agri Robot, has not yet appeared in print, but may be found on the web (Toyama 2009). It actuates motors that coincide with the knees, hips, shoulders and elbows according to spoken commands. Its main purpose is helping farmers.

These two exoskeletons, as well as HAL, show exactly how the limitations on weight and power result in augmentation of few DoFs while the shoulder girdle and spine remain immobile.

As for other types, there are several wearable arm exoskeletons that augment all DoFs of the human arms and shoulder girdle (Schiele & Van der Helm 2006) (Folgheraiter et al. 2009). These are for rehabilitation and haptics and require only small output torques. Using such structures to assist lifting, however, would result in larger and heavier devices.

The XOS exoskeleton, manufactured by Sarcos, also remains unpublished (Mirchandani 2008). This full body suit requires an external power source, but can provide powerful augmentation. The robot's arms only interact with the human at the end effector, thereby allowing the shoulder girdle to move as well.

Another type of exoskeletal devices consists of arms supported on a fixed base. The purpose of such devices differs, but despite the freedom regarding weight and power girdle motion has received limited attention (Perry & Rosen 2006) (Liszka 2006).

Lastly, pneumatic muscle actuators have been used in a full body (Tsagarakis & Caldwell 2003) and an upper body exoskeleton (Aida et al. 2009), and have been shown to provide the torque required for lifting. They work like muscles making them very compatible with humans. The main challenge, however, is to make a wearable power supply.

So far it can be concluded that, using current technology, being wearable and energetically autonomous cannot be combined with having all DoFs active and powerful enough to lift, e.g., patients. Critically, shoulder girdle motion has not been implemented in a full body exoskeleton, and spine motion has not received any attention at all.

1.2 Towards a New HAL

The current full body HAL suit, HAL-5, shown in Fig. 1a, consists of frames interconnected by power units, which each contain an electromotor and reduction gears, and are positioned directly next to the hip, knee, shoulder (flexion) and elbow joints of the wearer to assist his movements. Additional passive DoFs are located at each shoulder, upper arm, and ankle joint. The suit is powered by batteries.

The system is controlled according to the intentions of the wearer, which are obtained by measuring the bioelectric signal (BES) on the skin above the main flexor and extensor muscles associated with each augmented human joint. Each motor torque is calculated according to these signals.

It is expected that similar control techniques and actuators will be used in the new version. In addition, the wearer is assumed to be a healthy person.

Considering the found limitations and the aim to aid nurses, a new design approach for HAL should:

1) Achieve the most practicality given limited technology;
2) Enable handling of patients by nurses, by supporting the forces typically exerted by them.

The word practicality in the first goal implies "fitted for actual work or activities", and is considered necessary to achieve the most usefulness in human environments.

2 A DIFFERENT APPROACH

2.1 Challenges

These goals inevitably pose several specific challenges. Firstly, not being able to actuate a DoF poses the dilemma of creating either a passive DoF or a fixed structure instead. Passive means that the wearer will occasionally be required to exert a high degree of effort to handle heavy objects, whereas fixing reduces the human workspace and can result in high forces between the wearer and the robot.

When considering the practical usage of an exoskeleton it may be seen that both during daily tasks (Rosen et al. 2005) and during working (Vieira & Kumar 2004) gravity forces are the most prevalent. Although several exoskeletons specifically counter the forces of gravity during lifting (Suzuki et al. 2005) (Yamamoto, 2002) (Toyama, 2009), this focus also strongly limits the workspace by limiting various DoFs. Moreover, the loads should never be transferred from the suit to the wearer. E.g., as will be shown in section 4, the load supported by the suit may bear upon the wearer's body during walking. Some guarantee that the suit compensates gravity and transfers its weight and that of the carried load to the floor is necessary.

Considering patient-handling by nurses it can be seen that also pushing and pulling forces are prevalent, e.g., when turning a patient around in bed (Schibye et al., 2003). A useful exoskeleton will thus have to be able to support these forces as well.

Next, skin irritation around fastening equipment is a problem not often considered during design, but mostly revealed by experiments (Hidler & Wall 2005) (Colombo et al. 2000). Schiele and Van der Helm (2006) showed this partly arises from unavoidable misalignments between wearer and robot.

Regarding augmentation of the hands, which would be necessary for picking up heavy objects, it can be seen that only some fully actuated arms have wrist actuators (Schiele & Van der Helm 2006) (Folgheraiter et al. 2009) (Perry & Rosen 2006) whereas for the fingers there are only rehabilitation devices (Sasaki et al. 2004) (Mulas et al. 2005). Unfortunately, these devices also indicate that it is very difficult to augment fingers up to a practical load of around 25kg for one hand.

Lastly, shoulder girdle and spine motions, and thus their associated versatility, are very limited in current exoskeletons. One reason is that the shoulder girdle has two DoFs for each arm, and the spine has a total of three, requiring seven extra actuators, thereby almost doubling the amount on HAL.

2.2 A Human Practical Approach

Exoskeletal structures are typically designed using a machine approach, basing the design on only the range of motion (RoM) and torques of the human joints that they interact with. For W-EA robot suits it seems that with this approach current challenges cannot be overcome. On the other hand, the ways we use our bodies for work reveal characteristics that may provide unknown design opportunities.

In order to discover new solutions this paper posits that, although the number of postures and motions that may be achieved with the many DoFs our bodies provide is very large, we only use a limited subset of them in our daily lives and work because they are somehow optimal. If a robot suit can support this limited, practical set of postures and motions, then it may be considered sufficient.

To illustrate, it is possible for people to eat while maintaining their elbows at shoulder height. People generally avoid this because it is tiring. It is not practical. A practical design approach on the other hand goes beyond the machine-like characteristics of humans by considering what the wearer actually needs, wants, and does when wearing an exoskeleton.

What the wearer primarily needs is gravity compensation and the ability to move in ways that tasks may be performed as desired, without feeling the weight of the suit. Also, the suit must be aware of the wearer's intentions, as was realized with HAL's intention based control.

In particular the motions that are desirable or biomechanically optimal or motions otherwise used in practice, reveal solutions. E.g., the way an object is lifted reveals where and when augmentation is required. This is discussed further in the next section.

3 A SEMI-ACTIVE EXO-SPINE

3.1 Unified DoFs

The various muscles activated in the hips and back when lifting an object compose several DoFs. However, observing how they are activated, as shown in Fig. 1a, it can be seen that in the hips and back the moments are all generated in the same direction. They act as a single unified DoF.

Another example may be seen in Fig. 1b. By separating the trunk, pelvis, and legs, the interaction forces can be drawn schematically. This shows that during lifting - knowing that no other external forces are applied to the pelvis - the direction of the moments in the hips must always be the same as throughout the spine. This also holds when pushing forward or pulling backward. Pulling has the same effect as lifting, and pushing has the opposite effect. For ease of reference, this interdependence will be referred to as 'the lifting DoF'.

It also applies to adduction and abduction of the shoulder girdle. Abduction is connected to spine flexion, particularly when the body bends down to pick something up, as well as when pushing, and shoulder adduction is connected with spine extension during both lifting and pulling.

3.2 Semi-Active DoFs

It is possible to achieve a similar interdependency in the exoskeleton by using a semi-active DoF. This is a passive DoF driven by an active DoF.

Fig. 2 shows this concept schematically. Normally the stator of a HAL-5 motor moves an arm or leg segment and the axis is fixed to the exoskeleton frame. The axis may instead be connected to a pulley, and be allowed to rotate freely w.r.t. the frame using a bearing. This pulley then drives an otherwise passive joint through cables and a second pulley at this passive joint, much like a usual cable actuation system. The torques in the active and the

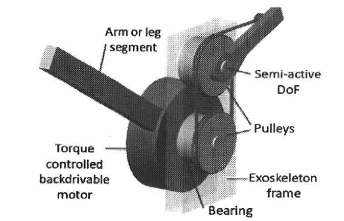


Figure 2. Placing bearings between the exoskeleton frame (transparent) and the axis of a hip (or other) motor while fixing the axis to a pulley, enables the pulley to drive a second joint that thus becomes semi-active.

semi-active DoFs are interrelated at any time, thereby creating the desired interdependency.

The motor is, as usual, torque controlled according to the BES of the wearer's muscles. When applying a semi-active DoF mechanism to actuate flexion of the exoskeleton spine, a torque controlled, back-drivable hip motor produces the same force balance in the exoskeleton as in the human lifting DoF. Adding abduction of the shoulder as a second semi-active DoF completes the robotic lifting DoF.

The moment in the human spine, however, decreases when the wearer bends, because the moment arm between the load and the spine, and the moment arm between the load and the hips change unequally. To achieve this effect with the robot, a four-link mechanism between each hip motor and the leg it drives may be used to increase the moment at the robot's legs when the legs are flexed, and thus relatively decrease the moment in the spine.

Because heavy objects are generally large, and are thus assumed to be held only in front of the wearer using both arms, the lifting DoF is applicable. Also, holding heavy objects on the side with one hand is unbalancing during walking and is not doable beyond normal human strength without hand augmentation, which, as mentioned, does not exist.

Considering this, and the ways we lift objects, it may also be seen that in a similar way elbow and wrist actuation can be connected into one unified, semi-active DoF, thereby simplifying design.

3.3 Exo-Spine Structure

In addition, it is possible to semi-actuate all three DoFs of the spine from both hips. First, just as the two hip moments in a human body act as one

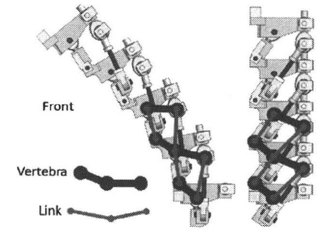
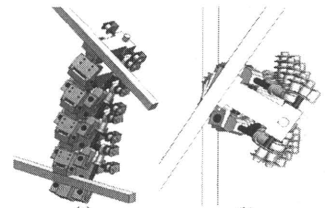


Figure 3. Spine structure composed of vertebrae and links, some overlaid by schematic equivalents, in fully bent and straight positions. It extends when bending forward to accommodate human spine flexion.



(a) (b)

Figure 4. Side bending (a) and rotation (b) (top view) of the spine structure. Beams were added for clarification.

moment on the trunk, the two axes of the two hip motors can be connected in order to let the total torque in this single axis act on the robot trunk.

Next, making sure that the exo-spine has a straight neutral position, any deviation should cause a moment that tends to restore the neutral position. This is applicable to each spinal DoF because when the wearer lifts something extra support is required in all directions in order to pick something up while being rotated, bend sideways or while using one hand. In practice, all three DoFs are connected into a unified DoF that tends to restore the neutral position.

A suitable spine-like mechanism is shown in Figs. 3 and 4. The details of the design are beyond the scope of this paper, but it can be seen that all three DoFs of the human spine are provided. Several vertebra-like segments and links are interconnected by ball joints, and two synthetic cables (not shown) connected to the axes of the hip motors pull the structure towards the neutral position. The cables continue from the spine upwards to support against girdle abduction during lifting and pulling.

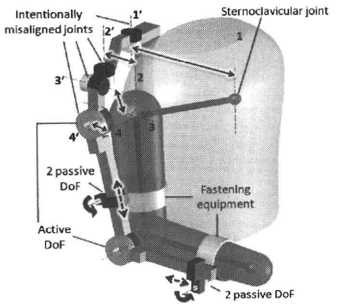


Figure 5. Schematic CAD model of the trunk and one arm. Robotic joints 1'- 4' do not align with the sternoclavicular joint (1) and the glenohumeral joint (2-4). Motions between the human and the robot at the fastening equipments are accommodated by extra passive DoF.

The connected axes of the hips are balanced in only one direction. In the other direction a torque-clutch locks the axes of the motors to the frame, depending on the direction of the combined torque produced by the motors and the cables.

Assuming 60Nm continuous torque per hip motor the exo-spine is analyzed using FEM to be strong enough to support 50kg at 24cm in front of the center of the wearer, which would relieve most of a nurse's load. Moreover, all forces applied to the wearer pull towards the neutral position and extension is blocked, making it safe to use. The exo-spine requires no extra actuators, provides gravity compensation, and supports pulling and pushing as required, utilizing practical human mechanics

4 OTHER SOLUTIONS

4.1 Intentional Misalignment

Two passive DoF added after each active arm joint as proposed by Schiele and Van der Helm (2006) not only facilitate the unavoidable misalignments between the robot and human joints, they also allows larger misalignments. Using this concept, offsetting the three robot joints at the shoulder w.r.t. the glenohumeral joint would create space for the wearer's shoulder girdle to move upward, without the need to actuate such a DoF. This increases the RoM of the arm, since raising the arm beyond about

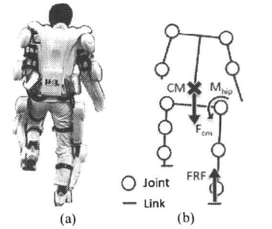


Figure 6. Single stance phase during stair climbing with robot suit HAL (a). A wireframe model (b) shows how the center of mass (CM) and the floor reaction force (FRF) create a moment at the hip joint of the stance leg, M_{hip}.

90 degrees involves upward motion of the girdle.

Fig. 5 illustrates this. Robot axes 1', 2', 3', and 4' have been misaligned intentionally w.r.t. their human counterparts. Axis 1' facilitates girdle abduction, supported by the same cables as the exo-spine. It has a large misalignment, but a small RoM, allowing it to function as desired. Due to the added passive DoFs the moment effectively put on the human arm for a constant motor torque differs by a few percent according to the posture. This is often disrupting for machines, but is not sufficient to influence the wearer of an exoskeleton.

4.2 Gravity Compensation

The human arm has a large workspace, but only some of the robot arm's DoFs can be active, and they must always be positioned such that they compensate gravity forces when required. In general, people often lift objects with their elbows kept down as much as possible, e.g. as farmers do (Nevala-Puranen 1995). Assuming this, an arrangement of passive and active DoFs as shown in Fig. 5 would let the output of each motor compensate gravity as much as possible. This is because the axes of the motors are always perpendicular to both the gravity forces as well as the line connecting the center of the motor to the point where the load is applied.

At the hips, forces acting from the back part of the suit are supported by the suit's legs. Standing on both legs is no problem, but when walking, during the single stance phase as shown in Fig. 6, a large moment, about 2Nm/kg, is developed around the hip of the stance leg. Because nurses on occasion need to walk when lifting, these moments should be supported. However, it is also desirable that the wearer be able to abduct the leg. Passive joints that only allow abduction of the legs would solve this.

5 DISCUSSION

Even with the proposed solutions the variety of motions that can be performed with HAL is still less than without, and limitations remain. Gravity compensation is limited to generic postures, some useful DoFs, such as inner rotation of the arm, are not augmented, and the full RoMs are not achieved.

Even so, in most working situations there are several postures available to the worker by which the task's goals can be achieved, and the wearer may adapt his motion to utilize postures for which HAL provides the most support. Since this is available in postures humans use extensively it is very likely that, although it should be confirmed by further research, at any time at least one good posture can be attained.

Therefore, HAL would still be valuable in a human environment and the proposed practical design approach thus achieved its goals. In addition, it is expected that further human characteristics may be exploited to simplify design or reveal solutions.

We believe that a similar practical focus may be applied to other fields where humans and machines meet cooperatively, such as rehabilitation, to yield new improvements. A practical approach could unveil solutions that enable patients to perform particularly those motions needed for daily activities.

ACKNOWLEDGEMENTS

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treating children of the PEDI scale in the group of children aged 4-9 and 10-15 is the biggest, while little or no improvement was found in the group of 16-18-years old. More significant dependencies between progress in rehabilitation and home environment in the group studied were not shown. Parents' education had little influence on the progress of rehabilitation.
Various methods ought to be used in the evaluation of progress in rehabilitation treatment of children with cerebral palsy. According to Maruszewski it is necessary to work with methods of objective measurement of the intensity of the disorders and the value of improvement obtained in the course of treatment. Zalewska and Sawa state that the search for optimum methods of diagnosing anomalies of development as well as effective methods of help determines the area of rehabilitation search. It is reflected in searching such ways and such methods of work which make it possible to realize the potential possibilities for development in them, despite the dysfunctions of the CNS.[7]
The studies demonstrate that the broadly understood comprehensive rehabilitation produces best results in the youngest children. Kruk-Lasocka shows the difficulties which a child with CP encounters when trying to master certain skills and habits of everyday life. It is evident that a child with CP finds him/herself in an undoubtedly unfavorable situation, caused by on the one hand, development limitation, and on the other hand, by unsuitable environmental conditions.[8, 9]
All in all, it needs to be stated that the earlier rehabilitation is introduced, the better effects may be obtained in the areas of mobility, self-care and social functions of children with CP. Through regular, long-term work of the therapeutic team and parents, contact with a peer group and a constant control of a child's development, it is possible for many children to adapt to everyday life.

Conclusion

1. Rehabilitation by means of the Conductive Education method by Petö allows to obtain an improvement in the areas of self-care, mobility and social functions. The biggest improvement was found in groups of children aged 4-9 and 10-15, while in the 16-20 age group, the improvement was the smallest.
2. In the group studied, no dependency between the progress in rehabilitation and the children's home environment, and little dependency between the progress in rehabilitation and parents' education were demonstrated.

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Use of a wearable robot — the hybrid assistive limb — to assist walking in a stroke patient: a case report

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Aim

The hybrid assistive limb (HAL) is a wearable robot that has been developped to physically support the wearer's daily activities and heavy work. The objective of this study was to clarify whether the HAL can assist the movement of paretic legs of stroke patients. We particularly focused on controlling the knee joint on the paretic side during the swing phase of walking.
This robot functions with a hybrid control system consisting of the following 2 algorithms: 1) voluntary control, which provides "power assistance" for the wearer's motion by receiving the bioelectrical signals arising from voluntary muscle activity, and 2) autonomous control, which provides the desired functional motion by receiving information other than that arising from muscle activity.[1] The HAL was originally developed as a two-leg system. Thereafter, a one-leg version was developed as a simpler system for persons with hemiparesis (Figure 1).

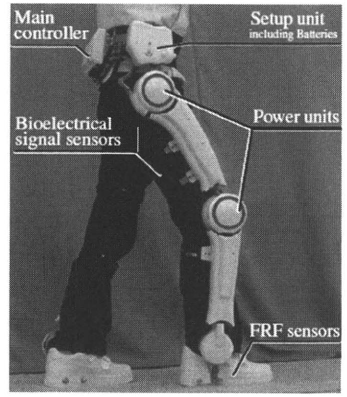

Figure 1. – Single-leg version of the HAL.
The HAL is composed of an exoskeleton, several power units, a main controller, set-up units that enable the user to adjust the activity of the robot, and the sensing system. The joints in the exoskeleton have 1 degree of freedom in the sagittal plane. Floor reaction force (FRF) sensors are attached to the insoles of the wearer's shoes.

Methods

We attempted to use the HAL on a patient with chronic stroke. The patient was a 59-year-old man who suffered from right hemiparesis due to cerebral hemorrhage that occurred 13 years ago. At the time of this experiment, the Brunnstrom recovery stage of his right lower limb was IV. With rehabilitation, he had regained the ability to walk with the help of a cane and an ankle-foot orthosis (AFO). The patient could hardly flex his right knee joint during the swing phase of walking. We attempted to assist this flexion movement of the knee joint by using the HAL. The patient provided informed consent for participation in the experiment. All the procedures employed were approved by the ethics committee of the relevant facilities.
Since the focus of this experiment was to control the knee joint movement, the ankle joint of the paretic leg was fixed at a right angle using an AFO. We also replaced the actuator of the hip joint to a free joint because the voluntary movement of the hip joint was so good that assistance was deemed unnecessary. These modifications partially reduced the weight of the HAL.

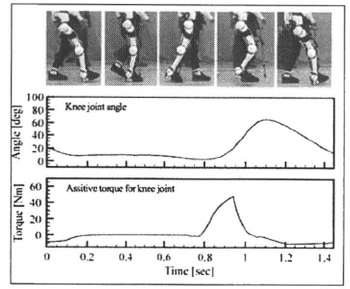

Figure 2. – Walking support with the HAL during one cycle.

It was difficult to apply voluntary control since the bioelectrical signals arising from muscle activity in the affected limb were distorted because of synergic movements and co-contraction of the antagonistic muscles. Therefore we set the HAL to function under autonomous control by sensing the weight shift of the wearer. This was achieved using floor reaction force (FRF) sensors, which were attached to the front and rear parts of the insoles of the patient's shoes. When the patient shifted his weight from the right leg to the left one during the double support phase in walking, the FRF at the rear of the left foot increased while that at the rear of the right foot decreased. The system was set so that the control of the right knee joint for the swing phase was triggered when the total FRF of the left foot became greater than that of the right foot. During the swing phase of the right leg, data obtained from able-bodied persons with regard to the angle and angular velocity during walking were applied as a reference to determine the pattern of proportional and derivative (PD) control of the knee joint. During the support phase, the motion control was also based on the PD controller, with a constant angle and a zero velocity applied as a reference.

Results

The patient could walk more than 10 m with the help of the HAL and a cane without experiencing any discomfort. No adverse effects were detected during this walking exercise. Figure 2 shows the data recorded during one walking cycle. It shows that the HAL generates the assistive torque in the direction of the flexion, and the knee flexion is achieved during the swing phase of the right leg. Figure 3 shows the knee and hip joint angles with and without the assistive action of the HAL during one walking cycle. We

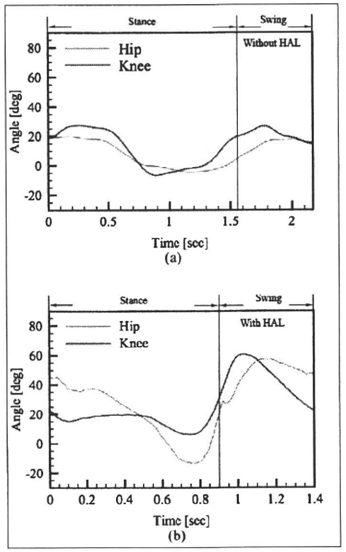

Figure 3. – The angles of the knee and hip jointt of the right side without the HAL (A) and with the HAL (B) during walking.

found that both angles measured during the swing phase increased when the HAL was used. Table I shows the length of the left and right strides and the time taken to walk 10 m with and without the HAL. The patient's right stride was longer and he walked faster with the HAL than without it.

Discussion

In most studies carried out on hemiparetic patients thus far, assistance to the knee joint has mainly been provided during the stance phase of the paretic limb in order to render greater stability while walking. Exoskeleton type robots which can actively assist the hip or knee joint during whole walking motion seem to be too large in size to be mobile at present.[2-4] We applied a single-leg version of the HAL to only 1 patient with hemiparesis as a preliminary study; nevertheless, we could confirm its effect during the swing phase of walking. Two issues should be investigated before the HAL can be employed for hemiparetic persons in clinical settings. First, our results revealed that although the subject experienced an improvement in his walking ability, the asymmetry in the stride length was in fact augmented. This implies that during the beginning phase of flexion wherein assistive torque was generated at the knee joint, the extensor muscles, including the rectus femoris, seemed to be contracting excessively. The assistive force that overcame this muscle contraction may have affected the hip joint motion because the rectus femoris originates proximal to the hip joint. The long-term effects of this type of control remain to be elucidated. The second issue that should be investigated further pertains to the weight of the HAL itself. Although the subject did not complain about the weight of the HAL during the short-distance walking exercise, the device produced an additional load of 7 kg. As a priority, ongoing research endeavors should aim to reduce the weight of such devices by using light materials and smaller components.

Table I. – Stride length and walking time with and without the HAL.

	Without HAL	With HAL
Left stride length (cm)	115	157
Right stride length (cm)	118	171
Walking time (10 m) (s)	33.2	13.4

Conclusion

In the present case, the HAL successfully assisted the movement of the knee joint on the paretic leg during the swing phase of walking and enabled the patient to walk without discomfort. Further investigations are required to clarify the long-term effects of such devices and to develop light-weight systems.

Acknowledgments.-This study was supported in part by the Grant-in-Aid for the Global COE Program on "Cybernics: fusion of human, machine, and information systems" at the University of Tsukuba.

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Voluntary Motion Support Control of Robot Suit HAL Triggered by Bioelectrical Signal for Hemiplegia

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***Abstract*—Our goal is to enhance the quality of life of patients with hemiplegia by means of an active motion support system that assists the impaired motion such as to make it as close as possible to the motion of an able bodied person. We have developed the Robot Suit HAL (Hybrid Assistive Limb) to actively support and enhance the human motor functions. The purpose of the research presented in this paper is to propose the required control method to support voluntarily motion using a trigger based on patient's bioelectrical signal. Clinical trials were conducted in order to investigate the effectiveness of the proposed control method. The first stage of the trials, described in this paper, involved the participation of one hemiplegic patient who is not able to bend his right knee. As a result, the motion support provided by the HAL moved the paralyzed knee joint according to his intention and improved the range of the subject's knee flexion. The first evaluation of the control method with one subject showed promising results for future trials to explore the effectiveness for a wide range of types of hemiplegia.**

I. INTRODUCTION

For patients with paralyzed extremities caused by a cerebral vascular disturbance like stroke, rehabilitation is conducted in order to recover motor function of the extremities during the acute and convalescence stages. For the chronic stage, rehabilitation is conducted in order to preserve the motor function as much as possible. However, it is not conducted therapeutically [1]. After the chronic stage, the patients who have motor paralysis will live their life using the residual function, making their daily life uncomfortable. For instance, patients who have difficulties from hemiplegia walk with a circumduction gait or a foot drop gait. These gaits are quite inconvenient to walk with in their daily life.

It is important that patients can live an independent life by compensating lost motor function using their residual function or an orthosis and cane, even if the compensated motion is different from an able-bodied person's motion. However, it would still be ideal for them to perform motions closely resembling an able-bodied person's motion. Here we suggest motion supports, by using a wearable robot.

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Wearable robots make it possible to support the impaired motion by creating external forces onto the paralyzed limb.

The motion supports are designed to be as closed as possible to the motion of an able-bodied person. This motor function compensation by using wearable robots will thereby enhance the quality of life of persons with paralyzed limb. Recent examples of wearable robots focused on the motor function compensation include the development of active orthoses and active prostheses [2]-[4]. Currently, one of the most important technical problems for these wearable robots is to obtain the user's motion intention. Moreover, applications of the wearable robot to patients with paralyzed lower limb have been hardly reported.

We have developed the HAL (Hybrid Assistive Limb) to enhance and support the human motor function [5], [6]. The HAL is a wearable robot designed for a wide range of applications, from welfare to heavy work support, and built in several versions (full body version two legs version and one leg version). In order to provide motion support according to the wearer's intention to move, we have developed a voluntary controller that uses the wearer's bioelectrical signals [7]. These signals are based on the myoelectric signal detected on the skin surface of the supporting muscles.

However, this voluntary controller is not well equipped to support the impaired motion of hemiplegia patients because the behavior of the myoelectric signal obtained from the paralyzed limb is different from that obtained from a healthy limb, and it is difficult to relate the myoelectric signal from the paralyzed limb to an adequate assistive torque.

The purpose of this research is to propose a voluntary control method that compensates the impaired motion of the paralyzed limb of hemiplegia patients and confirm the effectiveness of this method through experiments.

Section II describes the HAL systems. In section III a proposed controller are presented, followed by the experiments conducted with a person with hemiplegia patient (section IV). Section V contains the experimental results, and section VI discusses the effectiveness of the HAL in the context of hemiplegia. The paper's conclusion are presented in section VII.

II. HAL SYSTEMS

In this research we use the single leg version of the HAL, which was designed specifically for hemiplegia users. The external appearance of the HAL is showed in Fig. 1. The HAL basically consists of power units, the main controller,

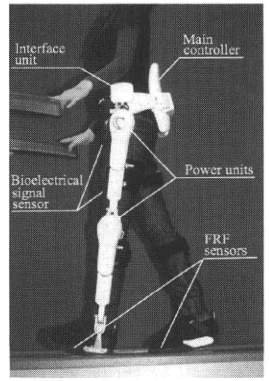

Fig. 1. Single Leg Version of the HAL

interface units (to allow the wearer to adjust the HAL behavior) and the sensing system. Furthermore, safety for the HAL and its users has been considered in the HAL's design.

The exoskeleton of the HAL system is an anthropomorphic structure designed to support the mechanical functions of the human lower body. It consists of a frame with active and free joints, and is attached to the user's hips and legs using cuffs and belts. The active joints attaching actuator (hip and knee) have one degree of freedom (DOF) in the sagittal plane. The free joints of the hip and ankle have 2 DOF in frontal and horizontal planes respectively. The required torques for motion support are generated by the power units of the active joints. Each unit integrates an actuator, a motor driver, a microprocessor and a communication interface. The motion support is achieved by transmitting the torques of the power units to the user's legs through the exoskeleton frames. One power unit is required for one joint. Actually, The HAL single leg version is thus equipped with two power units.

The control of the HAL system is ensured by the main controller. Its purpose is to control the power units, monitor the batteries, communicate with the system operator and supervise the safety of the users and the HAL. It modulates the assisting torque of each power unit in order to perform motion support for walking, standing up and sitting. Furthermore, it sends the sensors information and the system condition to the remote monitoring system, which provides a visual feedback to the operator and allows him/her to adjust the system parameters remotely.

The interface units contain an interface to adjust the parameters of the HAL, including the push buttons to tune the assistive gain. This interface allows the wearer to adjust the assist torque depending on his/her physical condition or desired level of comfort.

The HAL is equipped with a sensing system based on several types of sensors to detect the HAL's state as well as the wearer's muscle activities. Potentiometers are mounted on each joint of the HAL and used as angular sensors to measure the joint angles. The bioelectrical sensors are attached on the skin surface of the extensor and flexor muscles of the knee and the hip joint to detect their activity. Each shoe's insole contains two floor reaction force (FRF) sensors to measure the FRFs generated at the front and the rear of the foot (heel and ball areas). Furthermore, a gyro sensor and an acceleration sensor are mounted at HAL's body trunk to measure the absolute posture of the HAL based on the direction of gravity.

It is important to implement safety for the users and the HAL into the design of the HAL. As for safety mechanisms and functions, the HAL has mechanical joint limiters to remain within the human joint ranges of motion, and electrical limiters to limit the current range of each power unit. Furthermore, software safety components were integrated to detect malfunctions during motion support.

III. CONTROL METHOD

In this research, the motion support is focused on knee flexion support for disabled persons who have difficulties bending their knee due to hemiplegia but whose bioelectrical signals can still be detected from the impaired knee flexor. The control method proposed in this section mainly consists of command signal and torque generation.

A. Command signal

A command signal is used to trigger the motion support of the HAL. This signal can be controlled by the wearer's intention to move as follows. It is a binary signal obtained from the bioelectrical signal of knee flexor, BES_{kf}, and triggered when the wearer intends to bend the knee joint. The signal BES_{kf} is the signal detected by a bipolar electrode and rectified to full wave. In the following, the command signal, CS_{kf}, becomes 1 if the smoothed bioelectrical signal of knee flexor, $sBES_{kf}$, exceeds a threshold, BES_{kf_th}.

$$CS_{kf} = \begin{cases} 1, & \text{if } sBES_{kf} > BES_{kf\_th} \\ 0, & \text{otherwise} \end{cases} \tag{1}$$

The motion support for the knee flexion is triggered when CS_{kf} becomes 1. If $sBES_{kf}$ does not exceed T_{kf}, the command signal is kept at 0 and the motion support is not triggered.

B. Torque generation

When the HAL action has been triggered, an assistive torque is provided to the patient's knee joint. The torque generated by the HAL $\tau_{hal\_kf}(t)$ is described as follows,

$$\tau_{hal\_kf}(t) = \tau_{ka}(t) + \tau_{kd} + \tau_{kg} \tag{2}$$

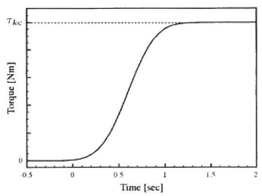

Fig. 2. Assitive torque pattern

The torque $\tau_{hal\_kf}(t)$ basically consists of three parts: 1) an assistive torque to drive the knee joint $\tau_{ka}(t)$, 2) a viscous torque to provide damping effectiveness τ_{kd} and 3) a gravity compensating torque τ_{kg}.

1) *Assistive torque*: $\tau_{ka}(t)$ is represented as follows,

$$\tau_{ka}(t) = \frac{\tau_{kc}}{2}\operatorname{erf}(kt - T) \tag{3}$$

where $\operatorname{erf}(x)$ is defined as,

$$\operatorname{erf}(x) = \frac{2}{\sqrt{\pi}}\int_0^x e^{-s^2}ds \tag{4}$$

This function is a kind of sigmoid-shaped function (error function), τ_{kc} a constant torque, k a parameter to adjust the rise time to reach τ_{kc} and T a parameter to change the time to begin to generate the assistive torque $\tau_{ka}(t)$. Fig. 2 shows the torque pattern generated by (3). Basically, the motion support torque τ_{kc} is kept constant during knee flexion support. Error function (4) modifies the start of τ_{kc} to let the assistive torque act smoothly on the wearer's lower limb until the assistive torque reaches τ_{kc}.

2) *Viscous torque*: τ_{kd} is described as follows,

$$\tau_{kd} = -k_d\dot{\theta}_k \tag{5}$$

where k_d is a damping coefficient and $\dot{\theta}_k$ is the angular velocity of the knee joint. This torque is generated to prevent high velocity motion and thereby maintain safety.

3) *Gravity compensating torque*: τ_{kg} is calculated as

$$\tau_{kg} = m_k l_{gk} g \sin(\theta_k - \theta_h) \tag{6}$$

where, as shown in Fig. 3, m_k is the mass of the lower leg and foot of the HAL and the wearer, l_{gk} the length from knee joint to the center of mass of the lower leg and foot of the HAL and wearer, g gravity and θ_k and θ_h the knee and hip

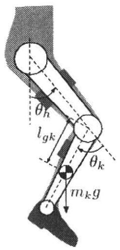

Fig. 3. Lower model of HAL and human

joint angle. By compensating gravity the feedforward torque can act uniformly on the wear's lower limb within the joint range of motion.

IV. EXPERIMENT

The following section presents the experiments conducted to validate the proposed control method by applying to a hemiplegic user. This section will introduce the characteristics of the trial subject and the experiment protocol.

The trial subject is a 60 year old male with right hemiplegia, resulting from a stroke. He has been diagnosed with Brunnstrom recovery stage IV. For locomotion in his daily

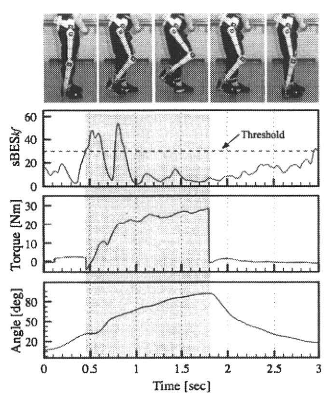

Fig. 4. Knee flexion support with the HAL during one cycle

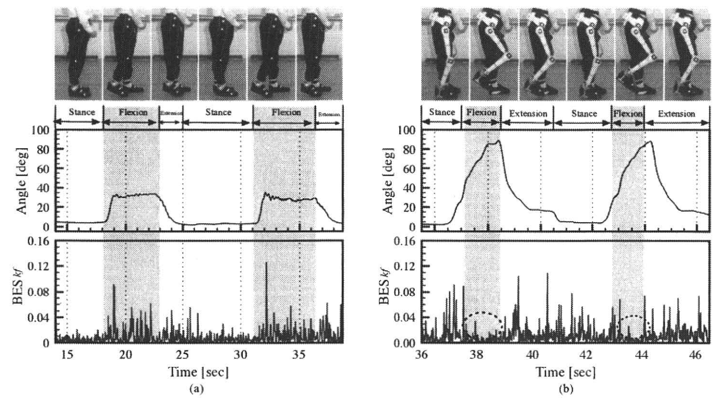

Fig. 5. Knee flexion during two cycles: (a) without the HAL, (b) with the HAL

life he wears an ankle orthosis, and walks using a cane. His walking is characterized by a circumduction gait, due to trouble to flex the right knee joint without flexing the right hip joint. We got informed consent before participating the experiment.

The trials were organized in one hour sessions, which were performed once a week, for four weeks. The experiment was performed in standing position. First, the subject stands evenly on both feet. After he shifts his weight completely to the left, the right knee flexion support is triggered by his intention, and consequently produced bioelectrical signals to bend his right knee. When the right knee joint is flexed to a preset angle the torque is released, he extends his right knee by himself and brings his weight by to his right leg. After that, he shifts his weight to the left again and motion support is performed once more.

In each trial, this step was repeated 7 times. Each time the subject has conducted a trial, some of the HAL's parameters were then adjusted to improve the knee flexion support. The parameters concerned are the threshold BES_{kf_th} which triggers the HAL action, and the amplitude of the assistive torque τ_{kc}. To ensure the subject's safety, additional support was provided to the subject. During each sessions, the subject held a handrail by his unaffected (left) hand.

In this experiment, the parameters k_d, m_k and l_{kg} in (5) and (6) are 10.0 Nm·s/rad, 4.73 kg, and 0.35 m respectively. The evaluation of the knee flexion support was made by comparing the data measured with and without using the HAL support. The comparison is based on the knee joint angle.

V. RESULTS

Fig. 4 shows data recorded during one cycle. The range of values of the smoothed bioelectrical signal change between trials depending on skin conditions and so on; the range of BES_{kf_th} in the figure is therefore arbitrary. The reference used for the knee joint angle is the value measured during the standing posture. Both the angles and torques are defined as positive during flexion. The parameters set during the experiments, τ_{kc}, k, and T in (3) in were 25 Nm, 15.0 and 0.02 sec respectively. It can be seen that when the bioelectrical signal of right knee flexor exceeds the threshold BES_{kf_th} which is 30, the HAL generates a positive assistive torque in the direction of the flexion and the subject's knee becomes flexed. After each parameter was fixed, the subject could repeat the knee flexion 7 times.

Fig. 5 shows the right knee joint angle and the bioelectrical signal of right knee flexor with and without assistive support, during two cycles out of a total of 7 times of knee flexion. The data shows that the right knee angle measured during flexion is larger when using the HAL. At the same time, the bioelectrical signal of right knee flexor during flexion while using the HAL (the encircled parts in Fig. 5 (a)) is smaller than that without the HAL (b).

VI. DISCUSSION

The HAL was used to assess the support provided to a person with hemiplegia. The motion support could be voluntarily provided by using a command signal based on the bioelectrical signal of the knee flexor. The range of the paralyzed knee flexion increased more than significantly. Furthermore, the use of the HAL allowed a decrease in the bioelectrical signal of the paralyzed knee flexor during knee flexion. This means that effort required for the paralyzed knee flexion decreased due to motion support from the HAL. As the result of experiments, we confirmed that the HAL could efficiently improved the knee flexion of the patient with hemiplegia.

Orthoses have been commonly used by persons with hemiplegia. Basically, standard orthoses assess paralyzed joints by through their stiffness property; it is impossible to move the joints voluntarily. However the HAL allows voluntarily support of the wearer's joints and actively compensates the motor function limitations caused by paralysis. Therefore the HAL appears to be more appropriate than orthoses as an assistive device for hemiplegia.

In the near future, the proposed method will be applied to the motion support of walking, standing up and sitting. Additionally, it would be a possible for a patient with hemiplegia to use this method as a rehabilitation tool. It has already been shown that voluntary activation is more effective than passive drive in the rehabilitation of paralyzed arms and wrists [8], [9]. We plan to conduct the rehabilitation of patients with paralyzed lowers limb by using this voluntary control method.

VII. CONCLUSIONS

In this research, a motion control method for the HAL robot suit was proposed in order to support the paralyzed leg of a hemiplegic wearer voluntarily. The bioelectrical signal detected on the skin surface around the muscle was used as the command signal, which is regarded as the wearer's intention to move, and a sigmoid-shaped torque pattern was adopted as an assistive torque to add the force to the wearer smoothly. To investigate the efficiency of the proposed method, the knee flexion support was assessed with one hemiplegic subject who has difficulties bending his right knee. As a result, the HAL allowed the subject to bend his knee voluntarily, which improved the knee flexion significantly, compared with the behavior without the HAL. This research suggests powered assistive devices may be more effective than orthoses for enabling natural motion of impaired limbs. The next step will be to explore the effectiveness of this proposed control method for a wide range of subjects showing different types of hemiplegia.

VIII. ACKNOWLEDGMENTS

This study was supported in part by the Grant-in-Aid for the Global COE Program on "Cybernics: fusion of human, machine, and information systems" at the University of Tsukuba.

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Development of Exo-Finger for Grasp-Assistance

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Abstract*—The most common outcome of stroke is hemiplegia and it is known that finger paralysis in particular tends to remain. These patients feel inconvenient in daily life because the grasping operation becomes difficult. On the other hand, not moving the paralyzed fingers for a long term will cause joint contracture. Therefore, daily use of the paralyzed finger is important to prevent joint contracture. To solve these problems, we propose applying a wearable assistive system that provides grasp-assistance for finger paralysis patients. The purpose of this research is to develop the "Exo-Finger" for grasp-assistance. The Exo-Finger is an exoskeleton system that assists the cylindrical grip that often appears during daily life. A wire is connected on the end point of the frame. When the motor pulls it, it is possible to extend the MP (Metacarpophalangeal), PIP (Proximal interphalangeal), and DIP (Distal interphalangeal) joints together. Elastics are used to assist flexion of the fingers. Moreover, we developed two kinds of control interfaces based on the state of the paralyzed finger. One is using a push switch operated by the healthy hand and another is using the small movement of the finger as a trigger. Through the experiment with a hand mock-up, grasping operation could be achieved as grasp-assistance with Exo-Finger. In conclusion, we developed the Exo-Finger and confirmed that the Exo-Finger would be used for grasp-assistance in daily life.

I. INTRODUCTION

Stoke is a serious brain disease. In Japan, several hundred thousand people develop stroke every year [1]. Stroke causes various aftereffects, and the most common is Hemiplegia. It is a motor dysfunction that makes patients unable to move the right or left side of their body. The condition of hemiplegia depends on the rehabilitation and level of brain injury. Especially, the part of the motor cortex responsible for finger movements tends to be damaged by stroke. It is known that finger dysfunction tends to remain. Moreover, the number of patients who can receive sufficient rehabilitation is restricted because of the restrictions of the rehabilitation period and shortage of therapists. Therefore, most hemiplegia patients live daily life with paralyzed fingers.

Once paralysis occurs in fingers, it becomes difficult for the patients to grasp an object. The grasping operation appears in all scenes of daily life. Thus, the loss of grasping function makes patients' daily life inconvenient. And it causes decrease of Activities of Daily Living (ADL) and Quality of Life (QOL).

Meanwhile, when the patients don't move their paralyzed finger for a long term, it causes joint contracture. It causes limitation of the range of joint motion, and it becomes an obstacle of daily life and rehabilitation. In order to prevent joint contracture, it is effective to use the paralyzed finger in daily life or to move it by external force. However, it is difficult for finger dysfunction patients to use the paralyzed finger. Additionally, when the joint contracture occurs, it becomes more difficult to use the paralyzed finger, and results in worsening of the joint contracture. Therefore, in order to prevent joint contracture, it is important to move the paralyzed finger by external force.

In short, the requirements for the patients are summarized as follows;

1) *Assisting the grasping operation of patients*
2) *Moving paralysis finger to prevent joint contracture*

These two things are important for finger paralysis patients. In order to actualize the requirements, we are focusing on a wearable assistive system to provide grasp-assistance for the paralyzed finger. By using a wearable assistive system, it is expected that increase of ADL and QOL by grasp-assistance and prevention of the joint contracture by daily use of the paralyzed finger. Additionally, daily use of the paralyzed finger promotes the neurorehabilitation [2]-[5] and there is a possibility that the paralysis improves.

For these patients, two types of assistive systems have been developed. First, Engen plastic hand orthosis [6] is a functional orthosis to assist grasping. This orthosis enables grasp-assistance by converting the wrist motion into the motion of the MP joint through a link rod. It is a lightweight orthosis because it doesn't require power such as an electrical motor. However, it is impossible to apply it for patients who cannot move their wrist, and the application is limited. Second, wearable type grasp-assistance systems using power sources such as electric motors and pneumatic actuators have been developed [7]-[10]. This system has possibility to use for the patients who cannot move their wrist. However most of those systems can assist only index and middle finger and the PIP and DIP joints are fixed. From the viewpoint of the prevention the joint contracture, it is important that the patient moves all his fingers on the paralyzed side [2]. Therefore, the wearable system that provides grasp-assistance for multiple fingers is required.

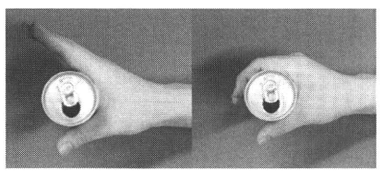

Fig.1. Cylindrical grip. Four fingers flex to fixed thumb together. Exo-Finger assists this grasping form.

The purpose of this research is to develop the "Exo-Finger" for grasp-assistance of finger paralysis patients. In this paper, we develop the hardware and the interface of the Exo-Finger and confirm the effectiveness of grasp-assistance by an experiment with a hand mock-up.

II. DESIGN OF EXO-FINGER

A. Target Patients

BrunnstromStage [11] is an evaluation method that stages the recovery of the motor function after stroke. This method is widely used to evaluate motor function of hemiplegia patients in Japanese hospitals. BrunnstromStage concerning finger motor dysfunction is as follows.

Stage1: Flaccidity
Stage2: Little or no active finger flexion
Stage3: Mass grasp, use of hook grasp but no release, no voluntary finger extension
Stage4: Lateral prehension, semivoluntary finger extension, with small range
Stage5: A variety of grasping operation awkwardly performed and with limited functional use, voluntary mass extension with variable range
Stage6: all prehensile types under control, full-range voluntary extension, individual finger movements present but less accurate.

After stroke, the condition of the paralyzed finger is stage 1 or stage 2. These patients cannot move their finger voluntarily at all. Many hemiplegia patients recover up to Stage 3 or Stage 4 by rehabilitation. However, the number of patients who recover up to stage 5 and stage 6 that can voluntarily extend their finger is limited. And in most cases, their recoveries stay in stage 3 and stage 4, and the paralyzed finger remains flexed. These patients can flex their finger voluntarily, but they cannot extend their fingers to grasp or release an object. As a result, the patients cannot grasp the object well. If the patient at Stage 3 and Stage 4 extends his finger by the external force, it is expected that the patients are able to grasp an object by their residual function. On the other hand, patients at Stage 1 and Stage 2 cannot move their finger. For these patients, it is possible to assist extension and flexion of the paralyzed finger with wearable system in order to prevent joint contracture. Patients at Stage 5 and Stage 6 are already able to perform the grasping operation. Their problems are to improve the dexterity of their fingers. Therefore we targeted patients in Stage1-4.

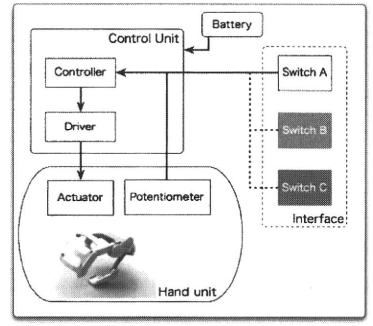

Fig.2. The Exo-Finger consists of hand unit, interface, and the control unit. The Exo-Finger can be used for various patients by using adequate interfaces.

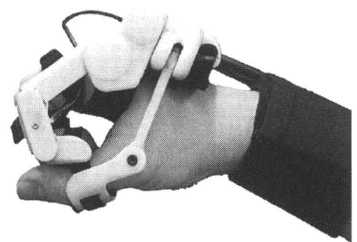

Fig.3. Main body of the Exo-Finger.

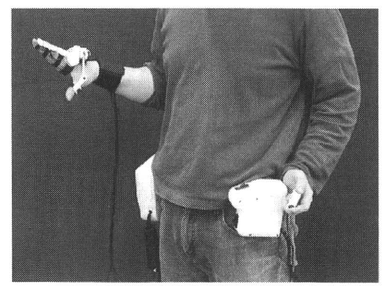

Fig.4. Appearance of whole system of Exo-Finger. The battery box and controller box are installed into the user's belt.

B. *Target Operation*

It is important that the patient uses all his fingers on the paralyzed side in order to prevent joint contracture. However, if the wearable system assists the motion of all fingers, the hardware of the system will become complicated and heavy. In this research, we selected a cylindrical grip (Fig.1) as the assistance operation to realize a simple and lightweight hardware. The cylindrical grip is a kind of grasping form that often appears in daily life [12]. In this grasping form, the four fingers (Index, middle, ring, and little finger) wrap the object while thumb is fixed. It is possible to move each joint of the four fingers by assisting cylindrical grip with the Exo-Finger. The grasp-assistance is indirectly performed by assisting extension of the four fingers as described in the foregoing paragraph.

C. *Required Specifications*

Exo-Finger is a wearable system that promotes daily use of the paralyzed finger by providing grasp-assistance. Therefore, it is required not only to perform grasp-assistance but also to wear the system without user's load. The following are listed as requirements of the Exo-Finger.

1) *Compact size and lightweight to be potable*
2) *Control interface corresponding to user's condition*
3) *Obtain enough range of motion*
4) *Assist paralyzed finger, and enable grasping*

The system of the Exo-Finger consists of the hardware (Hand unit), the interface and the control unit as shown in Fig.2. In this research, we develop a hardware and an interface that satisfy 1), 2). Next, it is verified that the hardware satisfies 3), 4) in an experiment with a hand mock-up.

D. *Hardware*

Figure.3 shows the main body of the Exo-Finger. It has the exoskeleton structure and the rotation axis in MP, PIP, and the DIP joint on the index finger side. And the thumb is fixed opposite to other fingers by the exoskeleton. The bridges are installed in the backside of the finger, and the frame on the forefinger side transmits power to the four finger through the bridge. The mechanical limiter is installed in the frame to avoid hyperextension for safety. An electric motor is installed on the back of the hand of the exoskeleton. Moreover, potentiometers are installed into MP, PIP, and the DIP joint on the index finger side, and it is possible to measure the joint angle. Velcro and rubber bands are used to attach the exoskeleton to the hand, it realizes the easy wearing. The finger frame is made of aluminum alloy, which has enough strength and lightness. The weight of the main body is about 230g. It is light enough compared with about 280g for the electric flexor hinge splint [7].

The appearance of wearing the whole system is shown in Fig.4. The control system can record, and monitor sensor information. A control box and a battery box are packaged individually. These can be installed in the user's belt with a hook. The user can move wearing the whole system.

The grasp-assistance mechanism is shown in Fig.5. In order

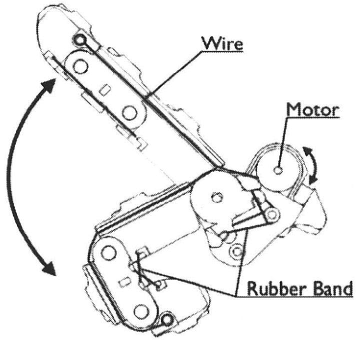

Fig.5. Grasp-assistance mechanism. Exo-Finger moves by tendon drive using wire and elastics. The electric motor pulls the wire, and then it extends all joint of the fingers together.

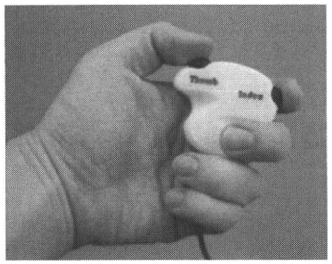

Fig.6. A control switch for patients who cannot extend fingers at all.

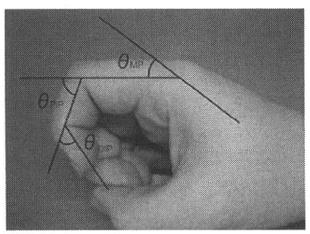

Fig.7. Joint angles

to realize a simple mechanism, the tendon drive with wire and elastic is adopted. The Exo-Finger assists the movement of each joint considering that the joint axis of the four fingers is about the same. The wire is connected on the end point of the frame. When the motor pulls the wire, the Exo-Finger extends the MP, PIP, and DIP joints together by a single degree of freedom. The elastics are used for assisting flexion, so it is possible to assist the grip force passively. By changing the strength of the elastic, it is possible to adjust the assisting grip force according to the grip force of the patients. Moreover, the finger fits the object and the motor dose not have to keep outputting power during grasping of the object.

E. *Interface*

It is required that the system has an interface based on the condition of the patients in order to promote daily use of the paralyzed finger. The Exo-Finger can be used for various patients by using adequate interfaces. In this research, we develop two basic interfaces based on BrunnstromStage for the evaluation of the Exo-Finger. Each interface is explained below.

Case1: BrunnstromStage1-3

Patients at BrunnstromStage1 to Stage3 cannot voluntarily extend their finger at all. For these patients, we prepared an interface with a small push switch (Fig.6). The user operates it using the healthy hand. It is a simple control method that switches the extension and the flexion with the push switch. The extension begins when the switch is pushed, and stops when the angle sensor detects reaching the limit of extension. When the switch is pushed again, the flexion begins. The finger comes in contact with the object during flexion, and the time, which the angle rate for each joint becomes 0, is judged as grasping the object, and operation stops. As a result, the Exo-Finger can grasp to fit the object. It is possible to control it without holding it, because the switch can be installed on the control box. In addition, this interface also can be used when the patient voluntarily exercises the finger in the early stage of rehabilitation.

Case2: BrunnstromStage4

It is important for patients to use the patient's residual function as much as possible to prevent contracture. The patients at BrunnstromStage4 can voluntary extend their finger a little. Therefore, we developed a control method using little extension of the finger as a trigger. The angles of the joints are shown in Fig.7, and the graph of the movement-trigger operation is shown in Fig.8. We use the angle rate of the PIP joint as a trigger. When the set threshold is exceeded, then the finger begins to extend. The extension of the finger is assumed to be the preliminary operation the grasping an object, and after the extension, flexion begins automatically after a set time. During flexion, the control method is same as in case1.

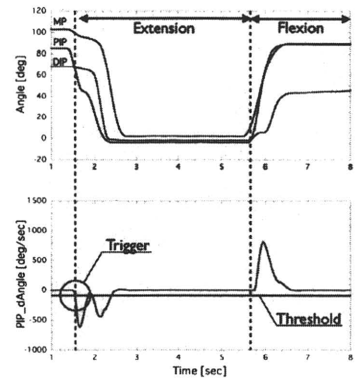

Fig.8. Graph of movement-trigger switching. Using little movement of PIP joint as a trigger.

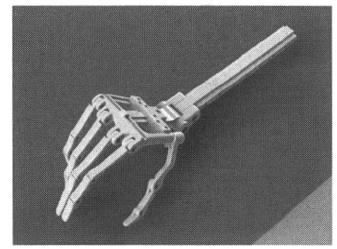

Fig.9. A frame of the hand mock-up.

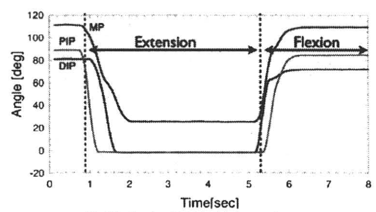

Fig.10. Graph of ROM with the mock-up.

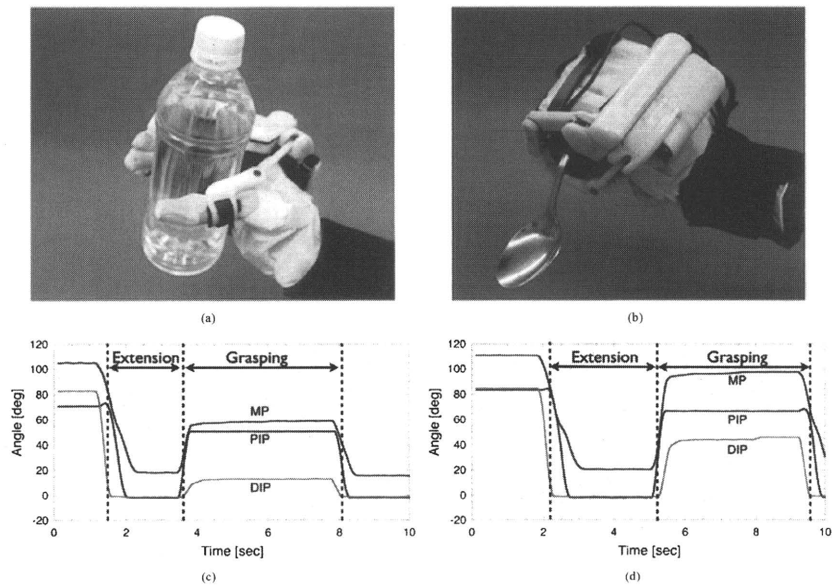

Fig.11. Grasping states and joint angles. Fig. (a), (c) is grasping a bottle. Fig (b), (d) is grasping a spoon. The angle in the graph shows that it is possible to grasp fitting to the bottle and the spoon.

III. Experiments & Results

We confirmed the range of motion (ROM) and effectiveness of the grasp-assistance as a fundamental experiment before applying the Exo-Finger to hemiplegia patients. We performed the experiments with a hand mock-up. The mock-up contains a frame shown in Fig.9. It is covered with sponge and contains three free joints in each finger. We performed the experiments by actual installing Exo-Finger to the mock-up. The push switch interface was used in the experiment for the control.

TABLE I
RANGE OF MOTION

	DIP ROM (deg)	PIP ROM (deg)	MP ROM (deg)
Human finger	0 – 80	0 – 100	-45 – 90
With mock-up	0 – 80	0 – 80	20 – 90

A. Range of Motion

First, we evaluated the joint's ROM with the mock-up. The graph of the operation is shown in Fig.10 and table.1 shows the ROM compared with human ROM. The flexion of the fingers is limited by the finger pulps, but it is enough to grasp the object. It is possible to perform the flexion and extension of a finger within one second. Moreover, The Exo-finger did not slip out from the mock-up while operating. Thus, it was confirmed that the Exo-Finger fits firmly on the mock-up.

B. Grasping Objects

To confirm the effectiveness of the grasp-assistance mechanism, we performed an experiment by actually grasping an object. A PET bottle and a spoon, which are often used in daily life, were selected as the grasping objects. The diameter of the PET bottle is 65mm, and weight is about 500g with water. The spoon was assumed to be a self-help device, and the axis diameter of 30mm was used. If the mock-up can grasp the objects with the Exo-finger, the hemiplegia patients would be able to grasp the objects because the remaining grip force of the patients is added to the Exo-Finger force output.

Fig.11 shows the images and the graphs of the experiment. The graph shows that it is possible to grasp fitting to the different size objects. Moreover the mock-up could grasp the bottle by passive assistance of the Exo-Finger. It would be possible to provide grasp-assistance by the Exo-Finger for the patients with low grip.

IV. Conclusion

In this research, we developed the Exo-Finger to realize grasp-assistance for hemiplegia patients. Exo-Finger is a lightweight system, and it enables to assist motion of each joint from index finger to small finger. Moreover, to evaluate the operation of the Exo-finger, we developed two kinds of basic control interfaces based on BrunnstromStage. It would be possible to expand the application of the Exo-Finger by using various interfaces based on condition of the patients. In the experiment with a hand mock-up, grasping a bottle and a spoon could be achieved as grasp-assistance with the Exo-Finger. We confirmed that the Exo-Finger would be able to use for grasp-assistance in daily life. As the next step, we plan to apply the Exo-Finger to grasp-assistance in daily life and rehabilitation.

Acknowledgment

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Development of HAL for Lumbar Support

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***Abstract*—Low back pain (LBP) is one of severe modern diseases due to the overload onto lumbar. Populations of 70% in the world suffer from LBP. The purpose of this study is to propose and develop the HAL for lumbar support in order to reduce lumbar load, and to verify the effect of the HAL on lumbar load reduction through an experiment at the task lifting a heavy load. Compressive force of lumbosacral disk and muscle strength of erector spinae can be represented as the function of lumbar moment. We proposed and developed the HAL for lumbar support in order to reduce the lumbar load. The HAL fixes movement of lumbar vertebrae and trunk flexion/extension motion is alternated to hip flexion/extension movement. The HAL reduces moment by supporting hip flexion/extension movement. In the experiment at the task lifting the heavy load, seven subjects lift the heavy load of 10 [kg] by using Stoop Lifting, first unsupported, then supported by the HAL. After the experiment, we executed three questionnaires as subjective assessment. From measurement results, it was confirmed that the HAL reduces the lumbar load corresponding to trunk flexion and dynamic motion. Results of subjective assessment showed that they could lift the heavy load easily. We proposed and developed the HAL for lumbar support in order to reduce the lumbar load, and verified the effect of the HAL on lumbar load reduction.**

I. INTRODUCTIONS

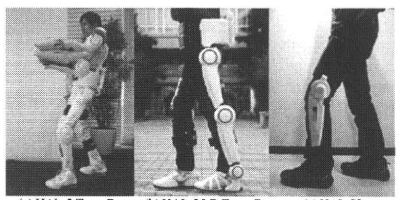
Low back pain (LBP) is one of severe modern diseases due to the overload onto the lumbar area such as vertebrae, disks and nerves. Populations of 70% in the world suffer from LBP [1]. In Japan, the survey by the Ministry of Health, Labour and Welfare has revealed that the number of patients with LBP account for 10% compared to the total population in Japan [2]. The pathogenesis of LBP is caused by physical demand on the lumbar during the keeping a static posture, the task lifting a heavy load, the spinning motion of a lower back, and so on [3]. These phenomena are found throughout the most of daily life, so that patients with LBP are forced to limit their daily activities and to decline the quality of life (QOL). LBP also brings on the serious problem in the field of healthcare. Caregivers usually help an elderly person in trunk flexed posture during the care of bathing, and transferring from a bed to a wheel chair [4]. From problems mentioned above, it is important to reduce the lumbar load of motions for not only patients but also healthy people.

An exoskeletal assistive system "Robot Suit HAL (Hybrid Assistive Limbs)" has been developed to enhance and support the wearer's physical activity [5]-[10]. In previous research, the HAL enhanced the wearer's ability to stand and walk by amplifying the wearer's own joint torque. It is verified and demonstrated that the HAL reduces the wearer's physical load. As the Figure 1 indicates, we already have developed several types of the HAL system by using these modularized technologies. There are full body type, upper body type, lower body type and single joint type [11]-[17]. It is important to realize that patients and healthy people perform motions without the lumbar load. The purpose of this paper is to propose and develop the HAL for lumbar support in order to reduce the lumbar load, and to verify the effect of the HAL on lumbar load reduction through an experiment at the task lifting a heavy load.

II. ANATOMICAL INSIGHT

A. Lumbar Structure

The spinal column of human and the two vertebrae segment are shown in Figure 2. The spinal column encases and protects the spinal code, supports the weight of the body. As the figure indicates, the spinal column is linear and has a symmetrical shape from the anterior view, and is a S-shaped curve from the lateral view. This complex curve is described as a lordotic curve. The curve comprises cervical lordosis, thoracic kyphosis, lumbar lordosis. In the human anatomy, the spinal column is usually consisting of 33 vertebrae. These are divided into four regions, cervical, thoracic, lumbar, and sacral. Cervical vertebrae consist of 7 vertebrae, denoted as C1-C7. Thoracic vertebrae consist of 12 vertebrae, denoted as T1-T12, these vertebrae increase in size and width from T1 to T12. Lumbar vertebrae consist of 5 vertebrae, denoted as L1-L5. Other vertebrae are sacral vertebrae that are denoted as S1-S5 and coccygeal vertebrae. Each vertebra is separated by disks. Vertebrae consist of the centrum and the vertebrae arch. The centrum and the disk are called anterior spine, assume a role of spinal support. Vertebrae arch is called posterior spine, assumes a role that adjust the function of movement. So, trunk flexed posture is based on flexion/extension movement of lumbar vertebrae and hip flexion/extension movement by rotation movement of pelvis centering on both hip joints. However, flexion/extension movement of lumbar vertebrae is accompanied by disk pressure.



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

Fig. 1. Robot Suit HALs. The HALs are developed to enhance and support physical ability of human being. HAL-5 Type-B is developed to support heavy work such as nursing care. HAL-5 LB Type-C can support complete paraplegic patients by estimating the wearer's intention based on a preliminary motion of their upper body and posture condition. HAL-SJ is very compact and supports the wearer's single joint.

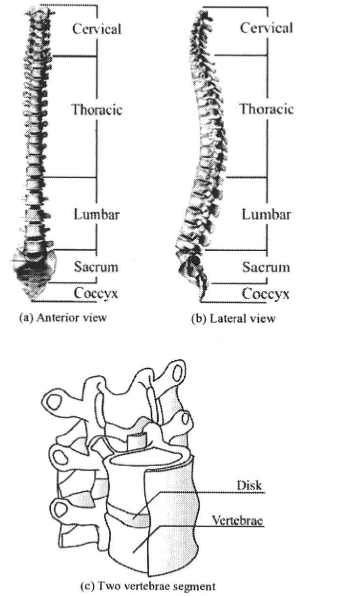


Fig. 2. Spinal column of human. The spinal column is the body's framework. The function of it is to support the body against the force of gravity, to protect the spinal code. The bones in the spine are called vertebrae. The spinal column starts at the base of the skull and continues to the pelvis. Disks between vertebrae absorb and distribute shock and keep the vertebrae from grinding together during movement.

B. Lumbar Load

Several studies have been conducted on lumbar load evaluation [18]-[19]. Compressive force is used as lumbar load evaluation index. Compressive force is the physical load onto disks, which is one of the causes of LBP. Measuring disk pressure directly is invasive. Thus, the estimation method which is non-invasive is used. The method estimates onto lumbosacral disk during static work. Lumbosacral disk lies between lumbar vertebra and sacral vertebra. The load is estimated based on posture and the weight of the heavy load etc. We consider trunk flexed posture as static work. The mechanical model of trunk flexed posture is described in Figure 3. Since articular surface of lumbosacral disk is inclined, compressive force and shear force is acting onto it. Compressive force causes disk injury and compression of nerves. Compressive force, F_c [N], is estimated by:

$$F_c = W_U \cdot g \cdot cos\theta_a + W_O \cdot g \cdot cos\theta_a - F_a + F_e \quad (1)$$

where W_U [kg] and W_O [kg] are the weight of upper body of the wearer and the heavy load, g [Nm/s^2] is gravitational acceleration, θ_a [deg] is the angle of articular surface of lumboscal disk, F_a [N] and F_e [N] are the muscle strength of erector spinae and abdominal muscle. In Equation (1), compressive force is decreased by activity of abdominal muscle, while increased by activity of erector spinae. These muscles contract to stabilize the spine, and erector spinae contracts to balance moment. Muscle strength of erector spinae, F_e [N], is calculated by:

$$F_e = \frac{M_{L5-S1} - F_a \cdot D_a}{D_e} \quad (2)$$

where D_a [m] and D_e [m] are moment arm of abdominal muscle and erector spinae. Muscle strength of erector spinae can be represented as the function of moment. In the same way, compressive force can be represented as the function of moment below. Lumbar moment, M [Nm], is calculated by:

$$F_e = f(M)$$
$$F_c = f(M) + C \quad (3)$$

$$M_{L5-S1} = W_U \cdot g \cdot G_U + W_O \cdot g \cdot G_O \quad (4)$$

where G_U [m] and G_O [m] are the distance to the center of gravity of upper body and the heavy load.

Judging from points above, the load that is applied on the disk and the muscle can be reduced by reducing moment. Lumbar moment is defined as the lumbar load in this research.

III. HAL FOR LUMBAR SUPPORT

A. System Configurations

We propose a method to reduce the lumbar load corresponding to trunk flexion and dynamic motion by using the HAL during the task lifting a heavy load. As mention before, trunk flexion/extension motion consists of flexion/extension movement of lumbar vertebrae and hip flexion/extension movement. When the HAL is developed, it is important to adjust the rotation axis of HAL to the wearer. However, it is difficult to adjust the rotation axis of the HAL to the rotation axis of lumbar vertebrae. In addition, movement of lumbar vertebrae is accompanied by disk

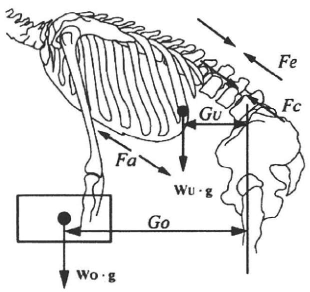


Fig. 3. Mechanical model of trunk flexed posture. W_U and W_O are the weight of the wearer's upper body and the heavy load, G_U and G_O are the distance to the center of gravity of the wearer's upper body and the heavy load, F_e and F_a are muscle strength of erector spinae and abdominal muscle, F_c is compressive force.

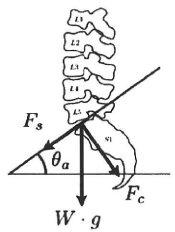


Fig. 4. Articular surface of lumbosacral. Articular surface is inclined by lumbar lordosis. θ_a in this figure is the angle of gradient. Vertical force is divided into two forces, which are compressive force and shear force. F_c and F_s in this figure are compressive force and shear force.

pressure. Thus, we try to limit movement of the lumbar vertebrae with a corset. That is, trunk flexion/extension motion is alternated to hip flexion/extension movement by limiting movement of lumbar vertebrae. Consequently, we have to adjust only the rotation axis of hip joint. It is likely to reduce the lumbar load corresponding to trunk flexion by supporting hip flexion/extension movement. In addition, it is expected that an increase in compressive force which corresponds to movement of lumbar vertebrae is inhibited by limiting the movement of lumbar vertebrae. As a conventional treatment of LBP, it is also hoped that the corset will increase intra-abdominal pressure. For reasons mentioned above, we proposed the method that reduces the lumbar load at trunk flexion/extension motion by using the HAL that supports hip flexion/extension movement.

Based on points above, we developed the HAL for lumbar support. Figure 5 and Figure 6 shows the system configuration of the HAL and the outline view. The HAL consists of power units, exoskeletal frames, sensors and controller. Exoskeletal frames design the framework of 1 joint and 2 links that have 1 degree of freedom on sagittal plane, and the rotation axis of the frame accords with the rotation axis of the wearer's hip flexion/extension. Also range of movement (ROM) is designed in flexion 120 [deg] and extension 15 [deg] based on ROM of the general ordinary person. Exoskeletal frames are fixed to the wearer's body with molded fastening equipments of lumbar and thigh. Lumbar mold assumes the role to limit movement of lumbar vertebrae. Power units are directly attached on both hip joints of the HAL to support hip flexion/extension movement. The actuators torque is transmitted from the HAL to the wearer's body through the mold fastening equipments. Potentiometers are attached to both hip joints to measure the relative angles. A triaxial accelerometer located in a control box to measure absolute angle of the wearer's trunk. When the person attempts to move, nerve signals are sent from the brain to the muscle via motoneuron. The bioelectrical signal (BES) is very week biosignals can be detected on the surface of the skin. BES sensors are attached to estimate the wearer's intension. A controller and batteries are attached on the wearer's back, and motor drivers and other electrical circuit are allocated on each power unit.

B. Control Algorithms

In order to support various types of people from healthy person to patients, we have also designed control algorithms specialized to wearers. One of the algorithms, that is, "Cybernic Voluntary Control (CVC)" controls the actuator torque of the HAL to augment joint torque of the wearer according to voluntary muscle activity that is estimated from the BES. The BES including myoelectricity is useful and reliable information to synchronize the motion support with the wearer's motion because the signals are measured just before corresponding muscle activities. CVC can be used for a healthy person and a physically challenged person who is able to detect biosignals, and support the wearer's motor function. The assist torque by using CVC, τ_{BES} [Nm], is estimated by:

$$\tau_{BES} = G_{assist}(K_f e_{fl} - K_{ex} e_{ex}) \tag{5}$$

where e_{fl} and e_{ex} are the BES of flexor and extensor muscle detected on the skin, K_{fl} and K_{ex} are transform coefficients of the BES to torque, and G_{assist} is assist gain that adjust the assist ratio.

Additionally, the HAL has also another control algorithm, that is, "Cybernic Autonomous Control (CAC)" supports a functional motion that is designed by the wearer. We used the CAC to support the wearer's weight for reducing lumbar load as a gravity compensation algorithm. The gravity compensation algorithm supports the wearer's weight so as to reduce moment caused by trunk flexion, because gravity applies the wearer. The assist torque by using CAC, τ_G [Nm], is calculated by:

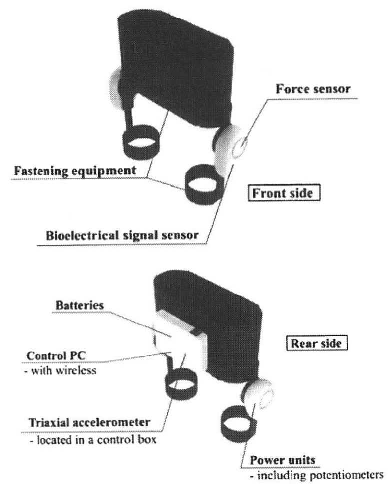


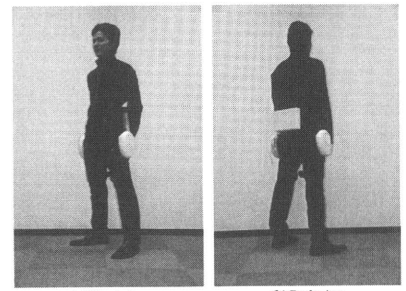
Fig. 5. System configurations of the HAL for lumbar support.

$$\tau_G = W_U \cdot g \cdot G_U \tag{6}$$

These algorithms generate the torque of each joint of the HAL. The HAL can control by both algorithms at the same time, the control algorithm is called "Hybrid Control". Depending on the wearer's condition, some parts of the wearer's body can be supported by CVC, while the other part can be supported by CAC. Trunk flexion/extension motion is caused the lumbar load on lumbar as for moment and joint torque. Thus, the HAL supports moment by CVC. At the same time, the HAL supports joint torque by CAC.

IV. EXPERIMENTS

The effect of the HAL on lumbar load reduction was verified by an experiment at the task lifting a heavy load. Lifting motion that is one of the pathogenesis of LBP is divided broadly into two techniques. There are Squat Lifting and Stoop Lifting in them. Squat Lifting is lifting motion by using knee flexion/extension movement, and Stoop Lifting is lifting motion by trunk flexion/extension motion. Therefore, Stoop Lifting is motion of moment that is larger than Squat lifting. In this research, we performed the experiment of lifting load by Stoop Lifting done daily and frequently. Target motion is motion to lift the heavy load by Stoop Lifting in the Figure 7. The wearer lifts the heavy load without HAL support, and lifts it with HAL support. The weight of the heavy load is 10 [kg]. Control algorithm of the



(a) Front view (b) Back view

Fig. 6. Outline view. The HAL supports the trunk flexion/extension motion by using power unit that are attached both hip joints. The HAL fits the wearer by being fixed the mold. And so, the rotation axis of the HAL matches to the rotation axis of the wearer's hip joint.

HAL is Hybrid Control. CVC estimates the wearer's intention, and supports joint torque caused by dynamic motion. To estimate the wearer's intention of trunk extended motion, the BES of erector spinae is measured in the Figure 8. CAC supports moment caused by trunk flexion onto lumbar. Subjects are seven adult men without LBP. After the experiment, we executed some questionnaire by the Likert Scale as subjective assessment of the wearer. Table I shows three questionnaires used by the experiment. The answer method is 5-point scale as follows. We rated subjective assessment by the mean value of each questionnaire's answer.

1. Strongly disagree
2. Disagree
3. Neither agree nor disagree
4. Agree
5. Strongly agree

V. RESULTS

Results of the trunk flexion angle and the HAL's output torque measurement in the Stoop Lifting appear in Figure 9. Desired torque is calculated using eq. (5) and (6). We can see from these graphs that Stoop Lifting is moment that is large because trunk flexion angle is large. As these graphs indicate,

TABLE I
QUESTIONNAIRES OF SUBJECTIVE ASSESSMENT

Q1. Did you feel the load by the weight of the HAL ?
Q2. Did you feel the target motion easy ?
Q3. Did you feel the pain by using HAL support ?