Comfortable Power Assist Control Method for Walking Aid by HAL-3

Kawamoto H.*, Sankai Y.**

*Sankai Lab, Doctoral Program in Engineering, University of Tsukuba
**Institute of Engineering Mechanics and Systems, University of Tsukuba
1-1-1 Tennodai, Tsukuba, 305-8573, JAPAN
*kawamoto@golem.kz.tsukuba.ac.jp **sankai@.kz.tsukuba.ac.jp

Abstract— **We have developed the power assist suit HAL (Hybrid Assistive Leg) which provide the self-walking aid for gait disorder persons or aged persons. The power assist was performed according to the operator's intention by using myoelectricity(EMG) signal as the primary command signal. However, there remains the problem that the operator senses the discomfort while the power assist is performed by EMG-based control. The scope of the present research is to propose the control methods to reduce the discomfort which the operator sense while the EMG-based power assist is performed in walking and standing up respectively; 1) the control method that the integrated motion between HAL-3 and the operator corresponds with the particularity of operator's musculo-skeletal system while in the power assist walking, 2)the predictive control method applying feedforward controller to remove the discomfort while in standing up. As the result, the power assist walking was performed corresponding to the joint kinematics of the normal walking without the assistance. Applying feedforward controller realized the quick response while in standing up power assist. These control methods enabled HAL-3 to realize the power assist that the operator sensed the reduce of discomfort.**

Keywords— **Power assist, Myoelectricity, Floor reaction force, Feedforward control**

I. Introduction

People with gait disorder can only move around by a wheelchair or by using a wheeled walker. Unfortunately, barriers such as bumps and steps restrict the area that these people have access to. Elderly people who are unable to walk without assistance may lose muscular strength in their legs and become bedridden. Caring for bedridden people entails a large physical burden, and in aging societies such as Japan, this has resulted in a growing perception of this issue as a social problem for the entire community to deal with. The most effective method ensuring that people do not become bedridden is to provide a way for them to be able to continue walking without assistance from a care-giver. In consideration of this problem, we have developed HAL(Hybrid Assistive Leg) series for such people. The HAL is a walking aid system which is capable of allowing the user to perform movements such as standing up, sitting down, and going up and down stairs[1][2][3][4]. The purposes of this research are to propose control methods to reduce the discomfort which the operator sensed while the EMG-based power assist is performed, and to realize an apparatus that enables power to be used for walking and standing up.

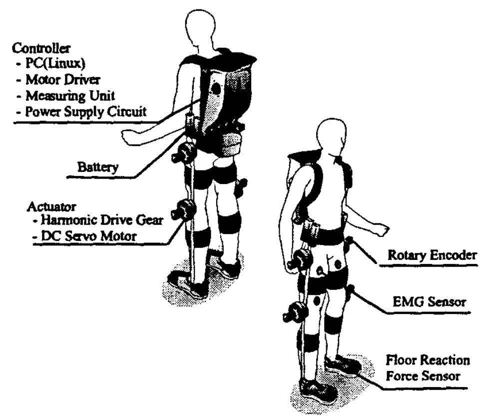

Fig. 1. Hybrid Assistive Leg (HAL-3) system.

II. HAL-3 System

HAL-3 system is composed of three main parts: skeleton and actuator, controller, and sensor. The schema of HAL-3 system is shown in Fig.1. The skeletal system of HAL-3 consists of exoskeletal frame. The exoskeletal frame is combined with the outside of the lower of limb, and transmit the assist force generated by the actuator to the lower of limb. The frame has joint at hip, knee, and foot respectively. The each joint has one degree of freedom, and the restriction is given at the joint mobile angle to ensure the safety of the person and to be correspondent to the movement of the human joint angle. The aluminum alloy and steel are used for the material of exoskeletal frame in consideration of lightness. The actuator system of HAL-3 provides the assist force for knee and hip joints. The actuator has DC-motor with the harmonic drive to generate the torques of each joint.

The control system of HAL-3 is mainly developed by considering the mobility because the field of activities of HAL-3 is presumed outdoors like corridors or stairs. So the compact type PC as the controller, driver circuits,

© 2002 IEEE SMC

Authorized licensed use limited to: NANKAI UNIVERSITY. Downloaded on March 27,2024 at 16:20:54 UTC from IEEE Xplore. Restrictions apply.

power supply, and measuring module are packed in the back pack. The real-time processing and the communication using the network are required in the control field. So the operating system of this PC is adapted as Linux which enables the measurement, the control, and the monitoring in real time. Wireless LAN(Local Area Network) card which has 11Mbps transmission speed, A/D(Analog to Digital) converter card which has 32ch(12bit resolution) inputs and D/A(Digital to Analog) which has 8ch outputs(12bit resolution) are selected respectively.

Sensor Systems of HAL-3 are used to detect HAL and operator's condition and estimate the assist force. The rotary encorder are prepared to measure the each joint angle, force sensors are installed in the front and rear sole of foot to measure the floor reaction force (FRF) sensor, and the myoelectricity sensors are attached on the surface of the skin of leg to estimate the muscle activity and the estimated torques for knee and hip joints.

III. Estimation of Assist Torque

Myoelectricity signal of a muscle relates the torque generated by the muscle. It is effective to estimate the assist force from myoelectricity signal. We need to decide the appropriate parameter relating myoelectricity signal to joint torque. So,we introduce a parameter calibration method using HAL-3.

A. Myoelectricity

Myoelectricity signals imply muscle activity, and are used as the estimation of joint torques. myoelectricity is action potential generated in a muscle as the command signal from motion control system of human is transmitted to the muscle through the motor nerves. The muscle contracts, after myoelectricity signal is generated. So myoelectricity signal can predict the beginning of generation of the muscle strength. The relationship between the joint torque and the processing myoelectricity signal in isometric contractions has been reported to be linear[6]. The joint torque would be estimated by the myoelectricity signal. Therefore, the appropriate assist torque can be estimated by using myoelectricity signals. The myoelectricity signals are measured through bipolar skin surface electrodes fixed to prepared skin over muscle. myoelectricity signal is amplified by 10^6 times, and filtered by using low pass filter cutting off at 500 Hz and high pass filter cutting off at 33 Hz to remove the effects by motion artifact. The filtered signal is transferred to PC through A/D converter. And imported signals are filtered again by low pass filter cutting off at 3 Hz to obtain the information of muscle force from spike signals[4].

B. Estimated torque

We find a joint torque from myoelectricity signals which are generated by the extensor and flexor respectively. The joint torques for knee and hip joints are estimated by using the following equations.

$$\hat{\tau}_{knee}(t) = K_1E_1(t) - K_2E_2(t) \qquad (1)$$

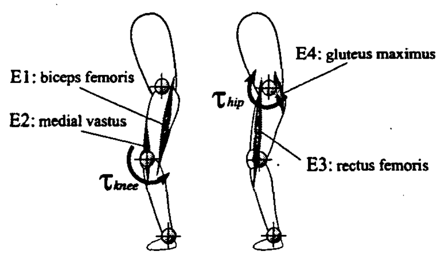

Fig. 2. Represented muscles in myoelectricity measurement. The joint torques are estimated based onthese myoelectricity signals.

$$\hat{\tau}_{hip}(t) = K_3E_3(t) - K_3E_3(t) \qquad (2)$$

where, $\hat{\tau}_{knee}$is estimated the torque for knee joint, $\hat{\tau}_{hip}$ is for hip joint. $E_1,\cdots,E_4$ are myoelectricity signals measured from the surface of the muscles shown in Fig.2. In equation (1) and (2), the term of the positive means extensor and negative means flexor. $K_1,\cdots,K_4$ are the parameters relating myoelectricity signals to the joint torques. These parameters are respectively identified by the calibration method.

C. Procedures for calibration

HAL-3 is used to determine the parameters $K_1,\cdots,K_4$. We assume that each joint torque is generated by only agonist. For instance, to determine the knee flexor parameter K_1, the torque $\tau_m(t)$ as the signal for the calibration is generated by HAL-3 knee actuator. The subject generates the knee joint torque $\tau_{fl}(t)$ in order to match with the added $\tau_m(t)$. The knee joint torque $\tau_{fl}(t)$ is equal to $\tau_m(t)$ generated by the knee actuator because the subject outputs $\tau_{fl}(t)$ to keep the knee joint angle constantly.

$$\tau_{fl}(t) = \tau_m(t) \qquad (3)$$

And the estimated torque calculated from myoelectricity of the flexor is represented as

$$\hat{\tau}_{fl}(t) = K_1E_1(t) \qquad (4)$$

The error $e(t)$ between the measured torque $\tau_{fl}(t)$ and the estimated torque $\tau_{fl}(t)$ is discretely represented as

$$e(k) = \tau_{fl}(k) - \hat{\tau}_{fl}(k) = \tau_m(k) - \hat{\tau}_{fl}(k) \qquad (5)$$

The performance function J can be expressed as

$$\begin{aligned} J = e^2(k) &= \sum_{k=0} (\tau_{fl} - \hat{\tau}_{fl})^2 \\ &= \sum_{k=0} (\tau_{fl} - K_1E_1)^2 \qquad (6) \end{aligned}$$

Authorized licensed use limited to: NANKAI UNIVERSITY. Downloaded on March 27,2024 at 16:20:54 UTC from IEEE Xplore. Restrictions apply.

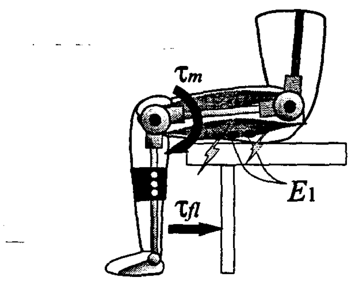

Fig. 3. System configuration of calibration to obtain the parameters by using HAL-3.

The performance function J can be minimized by setting its derivative with respect to K_1 equal to zero. This yields

$$\frac{dJ}{dK_1} = -2\sum \tau_m(k)E_1(k) + 2K_1 \sum {E_1}^2(k) = 0 \quad (7)$$

Therefore, K_1 can be expressed as

$$K_1 = \sum \tau_m(k)E_1(k) / \sum {E_1}^2(k) \quad (8)$$

The other parameters, $K_2, \cdots, K_4$ in equation(1) and (2) are calculated by the least squares method similarly.

D. *Experiment in calibration*

To obtain the parameters relating myoelectricity signals to the joint torque, we show the experimental protocols. The subject is normal 22 years old male. To measure the myoelectricity of knee flexor and extensor, the subject sits with the hip held at near $90[deg]$ to the upper body. The subject maintains the knee at near $90[deg]$ against the knee actuator torque $\tau_m(t)$(See Fig.3). In the same way, to measure the myoelectricity of the hip flexor and extensor, the subject keep upright standing posture. The subject maintains hip joint at $0[Nm]$ against the hip actuator torque $\tau_m(t)$. The reference torque $\tau_m(t)$ is made to increase from $8[Nm]$ to $32[Nm]$ in every $8[Nm]$ and is generated as a rectangular wave in the ten second period.

The parameters calculated for flexor and extensor of each joint are shown in Table I. For the right leg, the torque estimated using calculated parameters and the reference torque generated by actuator are shown in Fig.4. For the extensor and the flexor of each joint, the myoelectricity would be almost proportional to the torque generated by the actuator. Immediately the torque is generated by the actuator, myoelectricity increases in spike like. The subject relaxes, while the torque is not generated. As the torque $\tau_m(t)$ is formed, the joint angle is slightly moved over the desired angle which is maintained in calibration. The subject needs to generate the relatively large torque to bring back the desired angle. Therefore, the relatively large myoelectricity is produced. After this spike, myoelectricity is maintained at the constant value according to the actuator torque $\tau_m(t)$.

TABLE I
PARAMETER OF THE SYSTEM

Parameter	K_1	K_2	K_3	K_4
Right leg	3.1133	2.8412	3.6481	4.8901
Left leg	4.5742	2.4446	3.2213	6.9093

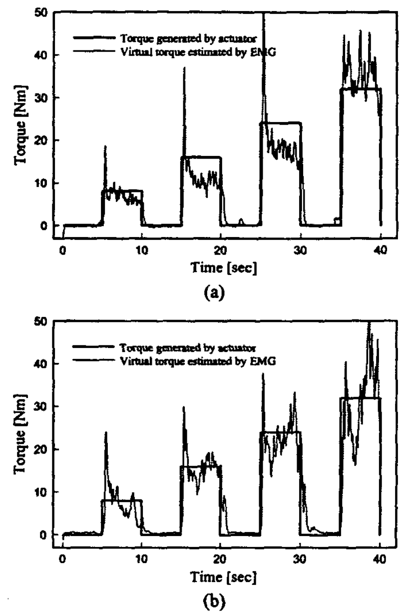

Fig. 4. Torque-myoelectricity relationship resulting from calibration. (a) Flexor of the right hip. (b)Flexor of the right knee.

E. *Experiment in power assist*

The experiments in walking and standing up are performed by using HAL-3. The assist torques are estimated according to myoelectricity signals generated in each motion.

E.1 Method

The subject is normal 22 years old male. To measure the myoelectricity signals, the skin surface electrodes are fixed to prepared skin over of the represented muscles (See Fig.2), and he wears HAL-3. The each parameters are obtained by the parameter calibration method which is shown in the preceding section. In the assistance of walking, the subject begins to walk from the initial condition of the standing posture. In standing up, the subject begins to stand up from the initial condition that the subject sits on the chair which is approximately 40cm high, and maintains the knee and hip joints at near $90[deg]$.

Authorized licensed use limited to: NANKAI UNIVERSITY. Downloaded on March 27,2024 at 16:20:54 UTC from IEEE Xplore. Restrictions apply.

E.2 Result and Consideration

Figure 5(a) and (b) show the estimated torques, the joint angles, and myoelectricity signals for hip and knee joint in walking. We divide the motion assisted by HAL-3 into phases corresponding to the particularities of the motion, and evaluate the assistance condition of each phase in comparison with normal walking.

E.2.a Walking. The motion of walking is mainly divided into two phases. Phase1 and Phase2. Phase1 is the swing phase that the foot gets away from the ground surface and the leg swings forward. Phase2 is the support phase that the foot contacts on the ground surface and the body is supported.

In Phase1, myoelectricity signals are mainly generated at the flexor of the hip and the extensor of the knee for the leg swings forward (Fig.5(a)). Then, the assist torques estimated by myoelectricity are generated, and the power assist performed the leg swings forward corresponding to the operator's intention.

In Phase2, myoelectricity signals are generated at the extensor and the flexor of the hip and the flexor of the knee. The extension of hip joints enables the body to move forward while the foot contacts on the ground surface(Fig.5(a)). In Fig.5(b), however, the hip joints are not enough extended in comparison with the normal walking(see dashed circles in hip joint angle). It is for this reason that the assist torques are influenced by myoelectricity signals generated at the antagonist muscles (the flexor of the hip). Then, the operator feels uncomfortable due to the restriction of the hip joint extension, while the power assist is performed in Phase2.

Therefore, the power assist in Phase1 can be performed according to the operator's intention, on the other hand, in Phase2 provides discomfort to the operator. As the next step of the power assist of walking, we need to develop the control method which avoids the influence of the antagonist muscle in Phase2. This control method is taken up in the next chapter.

E.2.b Standing up. Figure 6(a) shows the normal standing up without power assist and Fig. 6(b) shows the standing up while power assist is performed by HAL-3.

The motion of standing up from a chair is mainly divided into four phases. Phase0: sitting position, Phase1: the upper body is bent forward. Phase2: the upper body is lifted as the angles of hip joints attained to maximal value. Phase3: standing position. Especially, Phase1 and Phase2 are important to assist standing up.

In Phase1, myoelectricity signals are generated by the flexor and the extensor of the hip, and the extensor of the knee (Fig.6(a)). When the hip joints are flexed, the knee joints are slightly extended. The behavior of the assist torque(Fig.6(b)) is almost equal to the estimatd torque of the normal standing up. It means that the power assist is performed corresponding to the operator's intension. However, although HAL-3 generates the assist torque of knee joints , the operator feels uncomfortable at the beginning of the knee extension.

In Phase2, The assist torques are mainly generated at hip joints(Fig.6(b). The operator feels uncomfortable at the beginning of the hip extension. It is considered that the driving of the power assist is too late because the time lag occurs while the estimated torque is calculated from myoelectricity. In order to realize standing up aid comfortably, we need to perform the quick response power assist. This power assist method is taken up in the next chapter.

IV. Power Assist Control for Reduction of Discomfort

The floor reaction force(FRF) is the force acting to body from a floor. The motion of walking and standing up is performed based on FRF. FRF influences the behavior of the motion and the center of gravity of the body in walking and standing up. Therefore, It would be effective to analyse the behavior of FRF to detect the parts where the operator sensed the discomfort while the power assist is performed. In this chapter, we analyse the motion aid by power assist using EMG-based control, to propose a comfortable power assist methods using FRF sensor by considering some problems in the preceding chapter, and evaluate the effectiveness of these methods by experiments.

A. *Walking*

In walking aid, the operator felt uncomfortable because the extension of hip joint was prevented in Phase2. As shown in Fig.5(b), when the value of FRF is detected, the angle of hip decreases to less than 0[*deg*]. It means that hip joints are not extend. To realize the extension of hip joints in Phase2, we propose the control method which disable the influence of hip flexor myoelectricity signals in the equation (2) while FRF is detected. This control algorithm can be expressed as

$$\hat{\tau}_{hip}(t) = \begin{cases} K_3E_3(t) - K_4E_4(t) & (frf_r(t) \geq 0) \\ -K_4E_4(t) & (frf_r(t) < 0) \end{cases} \qquad (9)$$

where, frf_r is the value of the rear of FRF in the range of ±5[V]. The assist torques of hip joints are estimated from the equation(9). The experiments in walking are performed by using this control method. Figure5(c) shows the angle, the myoelectricity, the assist torque, and FRF while power assist of HAL-3 is performed in walking. It is clear that while the foot contacts on the ground surface, the extension of hip joints are realized. The behavior of hip joint angles in power assist walking controlled using FRF are almost equal to that in the normal walking. The operator sensed the reduction of discomfortable.

B. *Standing up*

The operator wearing HAL-3 feels a little discomfort in standing up aid because the time lag occurs while the estimated torque is calculated from myoelectricity. The assist method by the estimated torque using myoelectricity signals would be not adequate to assist standing up. It is considered that feedforward controller is adopted when human makes the agile motion like standing up. Feedforward controller is a kind of pattern generator. Therefore, it would be effective that the feed-

Authorized licensed use limited to: NANKAI UNIVERSITY. Downloaded on March 27,2024 at 16:20:54 UTC from IEEE Xplore. Restrictions apply.

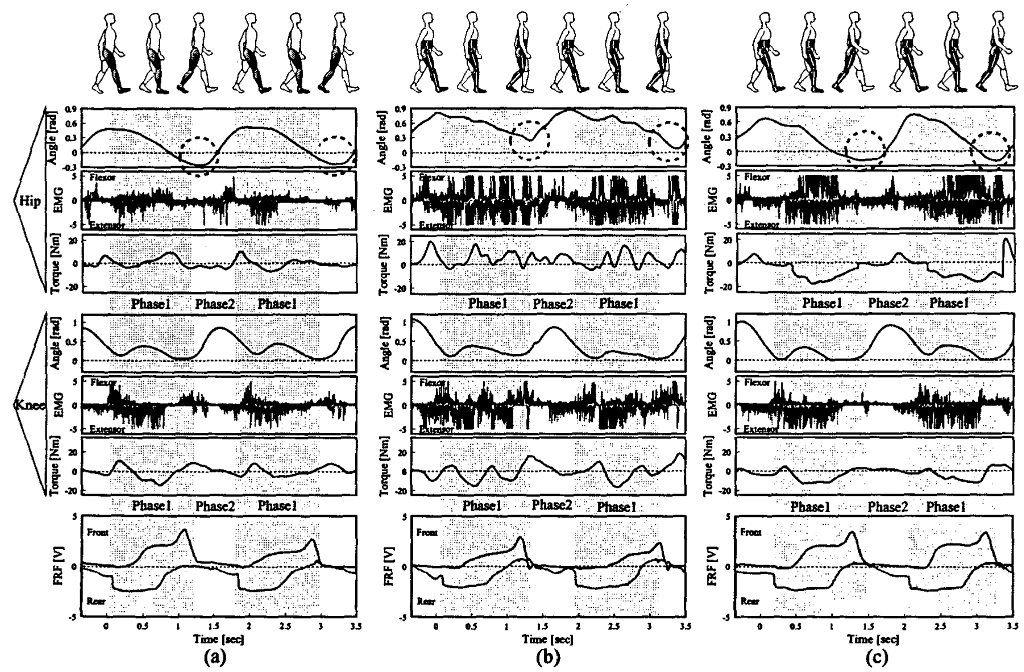

Fig. 5. Angle, myoelectricity signals and the estimated torque for left hip and knee joints respectively while power assist is performed in walking. (a) Normal walking without power assist (b)Walking by EMG-based power assist (c)Walking by EMG-based power assist using FRF.

forward controller is installed in order to realize quick response power assist like standing up aid.

We need to decide feedforward controller based on the analysis of the normal standing up motion in Fig.6(a) to utilize appropriate standing up aid. We note the behavior of FRF. The behavior of FRF can approximately predict the transition timing of each phase. Each transition timing is decided as follows from Fig.6(a).

Phase1 ⇒ Phase2

$$frf_r < V_{th}$$

Phase2 ⇒ Phase3

$$frf_r = V_{min}$$

Phase3 ⇒ Phase4

$$\theta_h < 0.3[rad] \cap \theta_k < 0.3[rad]$$

where, V_{th} represents the threshold value when the upper body begins to bent forward, V_{min} represents the minimum value of the rear part of FRF, and θ_h and θ_k are angle of hip and knee joints respectively. The feedforward controller in each Phase is represented as

Phase0(sitting position):

$$\tau_{ff\_knee} = 0[Nm] \qquad \tau_{ff\_hip} = 0[Nm]$$

Phase1(hip flexion, knee extension):

$$\tau_{ff\_knee} = -20[Nm] \qquad \tau_{ff\_hip} = 20[Nm]$$

Phase2(hip extension, knee extension):

$$\tau_{ff\_knee} = -20[Nm] \qquad \tau_{ff\_hip} = -20[Nm]$$

Phase3(standing position):

$$\tau_{ff\_knee} = 0[Nm] \qquad \tau_{ff\_hip} = 0[Nm]$$

where,τ_{ff_knee} is assist torque of knee joint and τ_{ff_hip} is assist torque of hip joint.

The experiments in standing up are performed by using the above method. Fig.6(c) shows the angle, the myoelectricity, the assist torque, and FRF while power assist of HAL-3 is performed in standing. The operator sensed the decrease of the discomfort. It is clear that the myoelectricity signals at hip joint decrease in Phase2 as compared with normal standing up. It means that the power assist is efficiently performed at Phase2 when the operator tends to feel discomfort in power assist using the only assist torque estimated by myoelectricity signals. This result means that the operator can stand up effectively. Therefore the feedforward controller would be regarded as essential to perform standing up aid.

V. Conclusion

To sophisticate the power assist control by HAL-3 for walking aid, we focused the reduction of the discomfort while in walking and standing up. In walking aid, We controlled the integrated kinematic motion of HAL-

Authorized licensed use limited to: NANKAI UNIVERSITY. Downloaded on March 27,2024 at 16:20:54 UTC from IEEE Xplore. Restrictions apply.

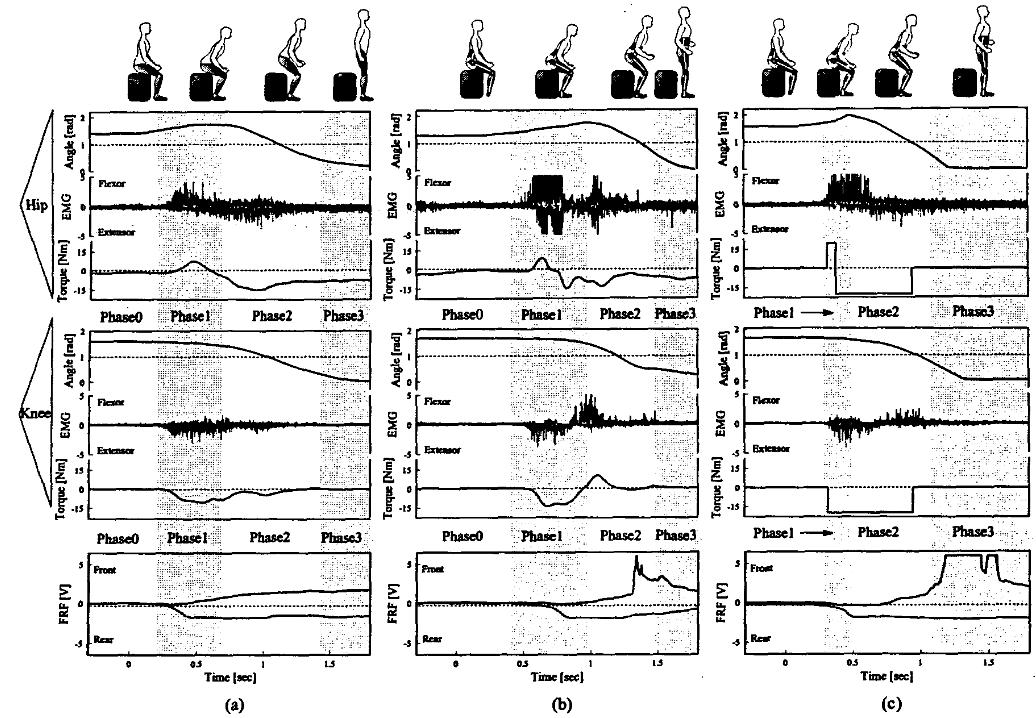

Fig. 6. Angle, myoelectricity signals and the estimated torque for left hip and knee joints respectively while power assist is performed in standing up. (a)Normal standing up without power assist (b)Standing up by EMG-based power assist (c)Standing up by feedforward control.

3 and the operator corresponding to the particularity of the operator's musculo-skeletal system. As a result, HAL-3 realized the motion corresponding to the kinematics of normal walking while in power assist walking. In standing up aid, We predicted the discomfort which appears while in standing up by using floor reaction force, and feedforward controller was installed to remove the discomfort, so that the power assist was performed with the quick response. The operator could obtain the comfortable power assist in walking and standing up aid. These control methods were regarded as essential to realize the comfortable power assist.

Acknowledgments

This research was partially supported by the Ministry of Education, Sience, Sports and Culture of Japan, Grant-in Aid for Scientific Research (A) and (B).

References

[1] Okamura J, Tanaka H, Sankai Y, "EMG-based Prototype Powered Assistive System for Walking Aid," *ASIAR'99*, Bangkok, Thai, pp. 229–234, 1999.

[2] NAKAI T, LEE S, KAWAMOTO H, SANKAI Y, "Development of Power Assistive Leg for Walking Aid using myoelectricity and Linux," *ASIAR2001*, Bangkok, Thai, pp. 295–299, 2001.

[3] Lee S, Sankai Y, "Power Assist Control for Walking Aid by HAL Based on Phase Sequence and myoelectricity," *ICCAS2001 International Conference on Control, Automation and system*, Jeju, Korea, pp. 353–357, 2001.

[4] Kasaoka K, Sankai Y, "Predictive Control Estimating Operator's Intention for Stepping-up Motion by Exo-Sckeleton Type Power Assist System HAL," *IROS2001*, Hawai, USA, pp. 1578–1583, 2001.

[5] Olney S, Winter D, "Prediction of Knee and Ankel moments of Force in Walking from myoelectricity and Kinematic Data," *Journal of Biomechnics*, vol. 18, No.1 pp. 9–20, 1985.

[6] Kawamoto H, Sankai Y, "EMG-based Hybrid Assistive Leg for Walking Aid using Feedforward Controller," *ICCAS2001 International Conference on Control, Automation and system*, Jeju, Korea, pp. 190–193, 2001.

[7] Kawamoto, H, Sankai Y, " Function Analysis Method of Human's Motion Control System," *IEEE International Conference on System, Man, and Cybernetics*, Nashvill, USA, pp. 3877–3882, 2000

Authorized licensed use limited to: NANKAI UNIVERSITY. Downloaded on March 27,2024 at 16:20:54 UTC from IEEE Xplore. Restrictions apply.