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Full paper

Power assist method based on Phase Sequence and muscle force condition for HAL

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Abstract—An exoskeleton robot can replace the wearer's motion function by operating the human's body. The purpose of this study is to propose a power assist method of walking, standing up and going up stairs based on autonomous motion of the exoskeleton robot suit, HAL (Hybrid assistive Limb), and verify the effectiveness of this method by experiment. In order to realize power assist of tasks (walking, standing up and going up stairs) autonomically, we used the Phase Sequence control which generates a task by transiting some simple basic motions called Phases. A task was divided into some Phases on the basis of the task performed by a normal person. The joint moving modes were categorized into active, passive and free modes according to the characteristic of the muscle force conditions. The autonomous motions which HAL generates in each Phase were designed corresponding to one of the categorized modes. The power assist experiments were performed by using the autonomous motion with a focus on the active mode. The experimental results showed that the wearer's muscle activation levels in each Phase were significantly reduced. With this, we confirmed the effectiveness of the proposed assist method.

Keywords: HAL; exoskeleton; power assist; phase sequence; myoelectricity signal.

1. INTRODUCTION

Exoskeleton robots have been studied in order to amplify human muscle strength. An exoskeleton consists of an external structure which covers the human body parts and has joint parts corresponding to those of the human body. Physical contact between the operator and the exoskeleton causes the integration of the operator and the exoskeleton. The exoskeleton directly provides mechanical power for the operator. Recently, exoskeletons for the human arms and their control methods have

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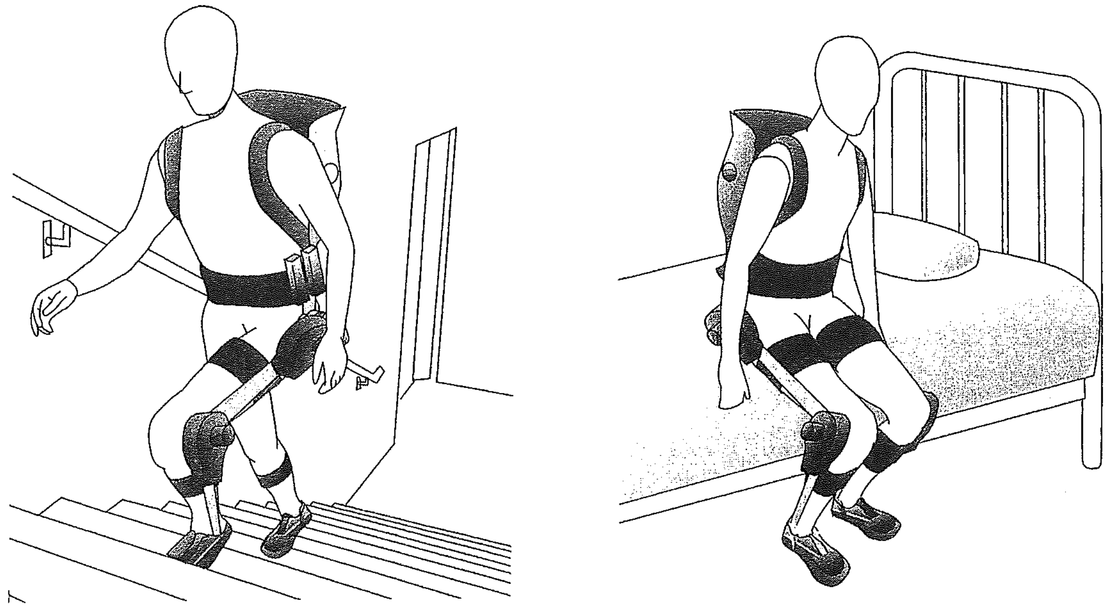


Figure 1. Images of power assist in daily movements.

been studied in order to extend human arm strength in order to reduce the burden imposed on the operator by an external load [1–4]. For example, if an operator wearing the exoskeleton manipulates an external load, the operator may feel 10% of the load while the exoskeleton carries 90% of the load. We have developed the exoskeleton robot suit HAL (Hybrid Assistive Limb) for the lower limb to provide power necessary to perform tasks such as walking, standing up, etc., as shown in Fig. 1. For example, if an operator cannot stand up because he generates only 70% of the muscle strength necessary for standing up by himself, he can stand up while the exoskeleton carries 30% of the muscle strength. We have proposed a power assist method using the joint torque, estimated based on the myoelectric signals that reflect the operator's intention [5–8]. Therefore, a common feature is that the operator manipulates the exoskeleton robot.

An autonomous motion generated by an exoskeleton robot moves the human's body directly. The human becomes a part of the exoskeleton system and the exoskeleton robot operates the human's body. The exoskeleton robot can take the place of the wearer's body function by performing human-like motion autonomically. If the exoskeleton robot has realized tasks such as walking, standing up, etc., the exoskeleton could be used as a functional aid apparatus for gait disorders in persons with spinal cord injury. Our research goal is to develop an autonomous method of motion of the exoskeleton robot in order to aid human leg motion.

When utilizing the exoskeleton robot as a human motion assist apparatus, the exoskeleton should provide motions like those of a human. We have developed the Phase Sequence method, using human motion characteristics, that enable humanoid robots to generate human-like motions [9, 10]. Phase Sequence is a method to generate tasks such as walking, standing up, etc., by transferring some prepared motion elements called Phases. A task is analyzed based on kinematic and biological information, and divided into a number of Phases according to specific motion intentions like 'swing the leg' or 'lift the body'. For each task, a sequence of Phases is transformed into the motion for the humanoid robot. As a result, the humanoid robot performs the task motions like a human. We attempted to adopt the Phase Sequence method to control the autonomous motions of the exoskeleton robot.

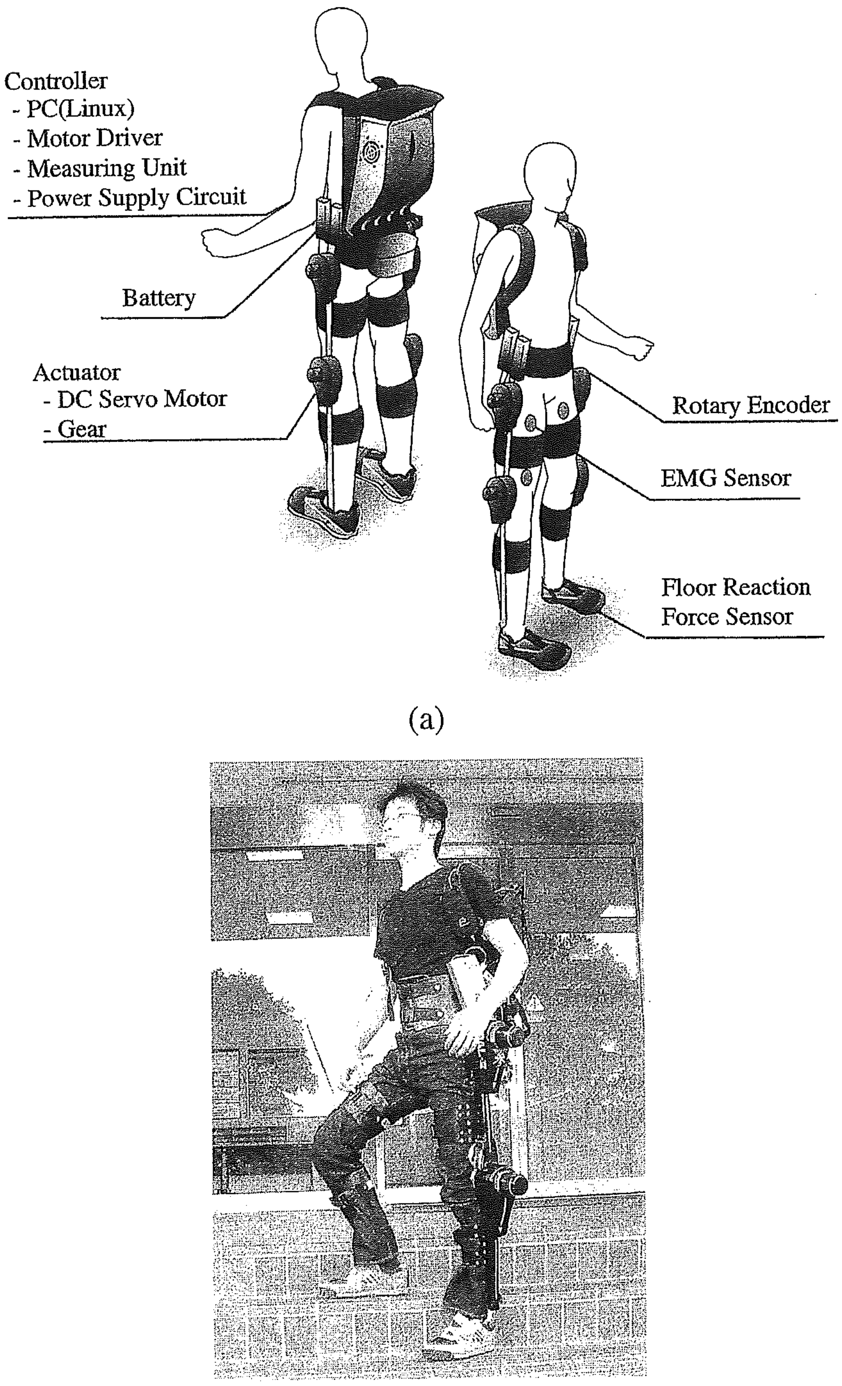
As the wearer and exoskeleton robot are combined into one integrated system, we consider that the motion properties generated by the exoskeleton robot accord to those performed by the wearer. The human realizes tasks using a variety of joint motions based on some modes of muscle activity. Thus, we consider that the exoskeleton robot should be used to generate motions corresponding to joint motions based on modes of muscle activity which a normal person performs. In this study, we propose the Phase Sequence method based on the modes of muscle activity for autonomous motion of the exoskeleton robot HAL. We divide the tasks of walking, standing up and going up stairs into Phases according to joint motion characteristics based on muscle activity. We realize the power assist for these tasks by performing and transforming Phases using HAL.

In Section 2, the details of the exoskeleton robot HAL-3 are described. In Section 3, the tasks of walking, standing up and going up stairs are divided into Phases, and Phase-shift timing for each task is determined. We construct the Phase Sequence control algorithm to generate assist motions. In Section 4, we describe the experiments on motion assists with Phase Sequence control and verify the effectiveness of this method. We give a brief conclusion in Section 5.

2. THE HAL SYSTEM

The HAL-3 system is composed of three main parts: a skeleton with an actuator, a controller and a sensor (Fig. 2).

The exoskeletal frame consists of a three-link, two-joint mechanism with the links corresponding to the hip, the thigh and the lower thigh, and the joints corresponding to the hip and the knee joints of the human body. Aluminum alloy and steel are used for the exoskeletal frame in consideration of lightness. The knee joint of the exoskeleton has 1 d.o.f. The human hip joint complex can be considered as a 3-d.o.f. joint. The exoskeleton, in its current mode, supports only the flexion–extension movement of the hip joint. The ankle joint has also 1 d.o.f., i.e. dorsiflexion–planterflexion. As shown in Fig. 3, the rigid sole is fitted with a fastener that is connected to the end of the lower thigh exoskeletal frame, which is provided with a swivel attachment. This swivel attachment corresponds to the ankle joint with 1 d.o.f. The joint limiters are equipped at the respective joints to prevent hyperextension of the hip and the knee joints. This exoskeleton system attaches to the hip, thigh, lower thigh and foot area of the body. At these areas (except the foot area), belts which are designed as shell garments of these areas worn by the wearer, are located at each link. At the foot area, the soles of the exoskeleton in which the wearer rides are provided. Since the human lower limb and the exoskeleton



(a)

(b)

Figure 2. The HAL-3 system. (a) Schematic overview. (b) System overview.

are mechanically linked, the movements of both legs of both the human and the exoskeleton are identical. The actuators of HAL-3 provide assist torque for the knee and hip joints. Each actuator has a DC motor with a harmonic drive to generate the assist torques at each joint. The ankle joints are not powered by a motor. If human walks quickly, the kick produced by the toe is used. We do not assume the

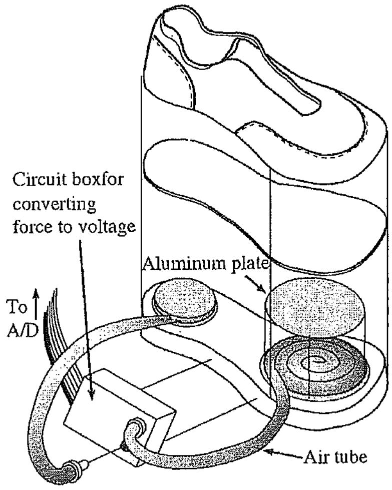


Figure 3. FRF sensor system.

assist for walking speed using the kick. We consider that the ankle joints play a important role in maintaining stability in the support phase. The spring mechanism is incorporated in the ankle joint of HAL-3. The total weight of the skeleton system with the actuators is about 15 kg. The wearer does not have to bear the weight load of the exoskeleton by riding on the soles of the exoskeleton, because the weight of the exoskeleton is transmitted to the floor, bypassing the soles.

Sensor systems are equipped on HAL-3 to detect the conditions of HAL and the wearer. Rotary encoders are used to measure the hip and knee joint angles. A floor reaction force (FRF) sensor is developed to measure the FRFs which are generated in the front and rear parts of the foot (ball and heel of foot). This sensor structure is shown in Fig. 3. It consists of a sole and electronic circuits parts. In the sole of the shoe, two coiled polyvinyl chloride tubes (inner diameter 3 mm) sandwiched by circular aluminum boards are installed. One end of the tube is connected to a solid-state pressure sensor attached on the electronic circuit. When the foot presses the tube, the air pressure in the tube changes. The change of air pressure is measured by the pressure sensor. With this, we can detect the FRF from the change of the air pressure. Myoelectric sensors are attached on the surface of the extensor and the flexor of the knee and the hip to detect muscle activity. Each myoelectric sensor consists of a bipolar electrode and a pre-amp, which reduce noise substantially. The myoelectric signals are first measured through the bipolar skin surface electrodes attached to the skin on top of the muscle and amplified by 106 times, and then filtered using a low-pass filter (with a cut-off frequency at 500 Hz) and a high-pass filter (with a cut-off frequency at 33 Hz) in the back pack to remove noise caused by motion artifacts.

The control system of HAL-3 is mainly developed to enhance mobility because the field of activity for of HAL-3 assistance is expected to be in outdoor activities. Thus, we designed a compact-type PC which is the controller, the motor drivers, the power supply for the PC and other circuits, EMG signals processing board and sensor interface boards to be packed in the back pack. The PC has a Celeron 566 Hz

CPU, a wireless LAN card (11 Mbps transmission speed), an A/D converter card which has a 32-channel (12-bit resolution) input and a D/A which has an 8-channel (12-bit resolution) output. Real-time processing and network communication are required for the control scheme. To make the development environment convenient, we adopt different operating systems for the measurement process and control process. RT-Linux is used for the measurement process. It is able to measure sensor information in real-time. On the other hand, we use Linux for the control process. Real-time processing can be achieved in practical use by modifying only one parameter of the Linux kernel source file. Thus, the control loop is executed in user mode by using this approach which almost guarantees a fixed control period [11]. To monitor sensor information with the remote controller in real-time, radio communication using UDP is utilized between the HAL controller and the remote controller.

3. POWER ASSIST METHOD

In this section we describe how to a divide series of motion into Phases and how to transit each Phase in order to apply the Phase Sequence method to the power assist system HAL-3.

3.1. Muscle condition

In the various tasks (walking, standing up, etc.), each muscle generates force mainly based on threes mode of muscle force conditions, i.e. active, passive and free, depending on the direction of muscle length contraction and external forces as shown in Fig. 4. In the case of the rectus femoris, the muscle length is shortened as its contractive force is generated in the active mode (Fig. 4a). This is mainly based

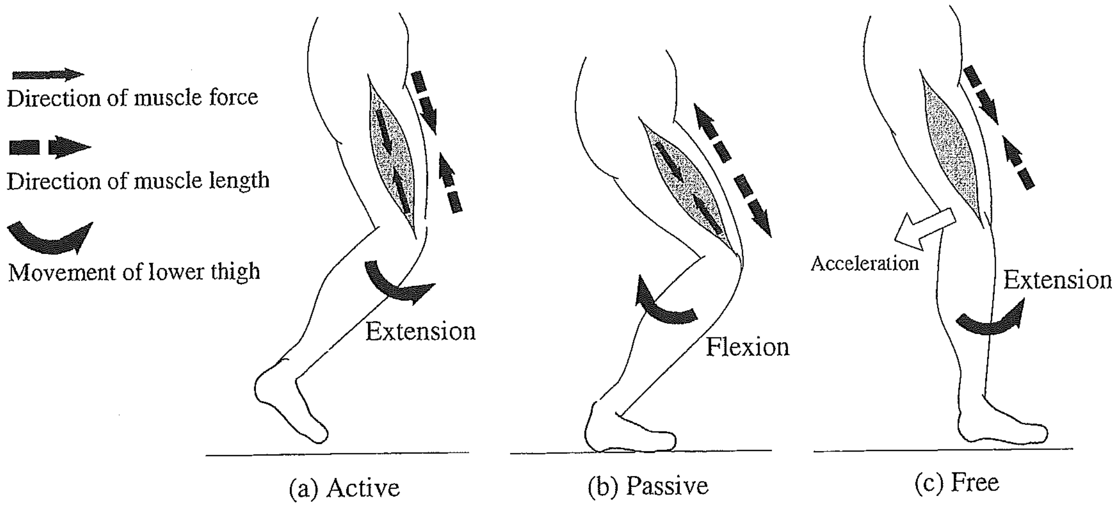


Figure 4. Muscle force mode depending on the relationship between the direction of muscle force and the direction of muscle length contraction.

upon the contractile element. In the passive mode (Fig. 4b), the muscle length is lengthened as its contractive force is generated. These muscles play the role of an viscoelastic element. In the free mode (Fig. 4c), the muscle contractive force is not generated. The movement of the lower thigh is generated by the force caused by knee acceleration or gravity. As the result, the knee joint behaves like a free joint. It should be effective for the power assist to generate assist motions corresponding to the wearer's muscle conditions.

3.2. Phase division

We analyzed the motion of a normal person during each task (walking, standing up and going up stairs) to divide it into some Phases according to the three modes based on muscle activity and direction of muscle length. The joint angle is assumed to be proportional to the muscle length. The muscle activities are estimated from the behavior of myoelectric signals of the flexor and the extensor. The subject was a normal 28-year-old male.

3.2.1. Walking. Figure 5a shows the joint angles and myoelectric signals for the hip and knee joint, and FRFs in the front and rear parts of the sole of the feet while walking. Each joint angle is set as 0 rad in the standing posture. Its positive and negative direction indicate flexion and extension, respectively. A positive sign of the myoelectricity corresponds to the flexor muscles and a negative sign corresponds to the extensor muscles. The activation level of the myoelectricity is represented in the range of ±5 V. The motion of walking is mainly divided into two phases — the support phase and the swing phase. The swing phase is the behavior where the foot leaves the ground surface and the leg swings forward. The support phase is the behavior where the foot stays in contact with the ground surface and the body is supported by the leg. In the swing phase (Phase 1), when the hip joint is bent, the myoelectric signals at the flexor of the hip are generated. The hip flexor works in active mode. At the same time, the knee joint is bent from the extension position and is extended after that. During the swing period the myoelectric signals at the flexor and the extensor of the knee joint are generated slightly. It is considered that the lower thigh is forced to move by the inertial force generated by the thigh. Therefore, the knee joint works in the free joint mode. In the support phase (Phase 2), when the hip joint is extended, the myoelectric signals at the extensor of the hip are barely generated. The hip extensor works in active mode. The knee joint is slightly bent from the extension position and is extended after that. The myoelectric signals at the extensor of the knee joint are largely generated when the knee joint is bent. The antagonist muscle (the extensor) performs lengthening contraction to absorb the shock to the knee joint from upper body when the foot makes contact with the ground surface. Therefore, the knee joint would work in passive mode.

3.2.2. Standing up. Figure 5b shows the joint angles and myoelectric signals for the right hip and knee joints, and FRFs in the front and rear parts of the right

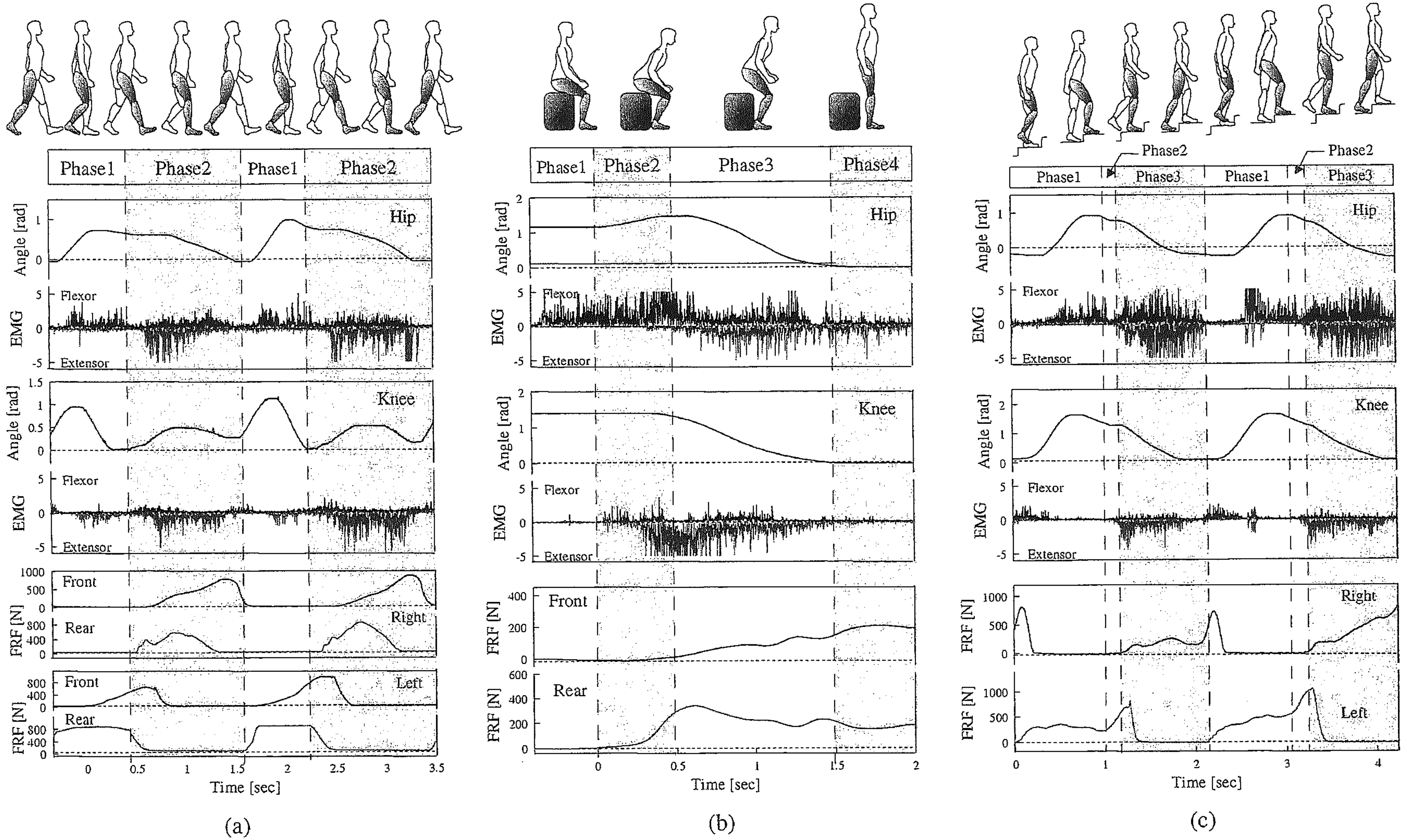


Figure 5. Analysis of a normal series of motions: (a) walking, (b) standing up and (c) going up stairs.

foot while standing up. The subject begins to stand up from the initial condition that the subject sits on a chair which is approximately 40 cm high, and maintains the knee and hip joints at near 1.5 rad. The motion of standing up from a chair is mainly divided into four phases. Phase 1 is the sitting position. Phase 2 is when the upper body is bent forward. Phase 3 is when the upper body is lifted as the angles of the hip joints attained the maximal value. Phase 4 is the standing position. Phases 2 and 3 are especially important to assist standing up. We analyze the muscle condition from myoelectric signals on Phases 2 and 3. In Phase 2, myoelectric signals are generated by the flexor of the hip and the extensor of the knee; when the hip joint is flexed, the knee joint is slightly extended. In Phase 3, when the hip and knee joints are extended, myoelectric signals are generated by the extensors of both joints. Because each muscle performs shortening contraction in Phases 2 and 3, both joints work in the active mode in each Phase.

3.2.3. Going up stairs. Figure 5c shows the joint angles and myoelectric signals for the right hip and knee joints, and FRFs in the front part of the both feet while going up stairs. The subject goes up stairs that are 15 cm high and 30 cm wide. The motion of going up stairs is mainly divided into three Phases. Phase 1 is the behavior that the foot lifts from the stair surface and the leg is lifted up. Phase 2 is the behavior that the leg is slightly lowered in order to establish contact of the foot with the stair surface, after the foot is lifted up above the stair surface in Phase 1. Phase 3 is the behavior that the foot contacts with the stair surface and the body is lifted up. In Phase 1, when the hip joint is bent, the myoelectric signals at the flexor of the hip are generated. Thus, the hip flexor works in active mode. On the other hand, when the knee joint is bent, the myoelectric signals at the flexor and the extensor of the knee joint are generated slightly. The knee joint is bent by lifting the thigh without moving the lower thigh. The flexor and the extensor of the knee joint do not act. Therefore, the knee joint works in the free joint mode. In Phase 2, when the hip and knee joints are extended slightly, the myoelectric signals at the flexor and the extensor of both joints are generated slightly. Each joint works in free mode. In Phase 3, when the hip and knee joints are extended, the myoelectric signals at the extensor of the hip and knee are generated. Each extensor works in the active mode. As explained above, we divide the tasks (walking, standing up and going up stairs) into Phases. Figure 6 shows the joint part, direction and dynamic mode in each Phase for walking, standing up and going up stairs.

3.3. Phase-shift timing

To realize power assist by using Phase Sequence, the prepared Phases have to smoothly transit. The Phase-shift timing needs to reflect the wearer's intention. If the Phase generated by HAL does not accord to the Phase which the wearer intends, HAL may provide unnecessary load to the wearer and the wearer would feel uncomfortable. In this section, we determine Phase-shift timing on the basis of the motion of a normal person. Most motions of the lower limbs of a person

(a)

	Phase 1		Phase 2	
Joint	Direction	Mode	Direction	Mode
Hip	Flexion	Active	Extension	Active
Knee	Flexion ↓ Extension	Free	Flexion ↓ Extension	Passive

(b)

	Phase 1		Phase 2		Phase 3		Phase 4	
Joint	Direction	Mode	Direction	Mode	Direction	Mode	Direction	Mode
Hip	—	Static	Flexion	Active	Extension	Active	—	Static
Knee	—	Static	Extension	Active	Extension	Active	—	Static

(c)

	Phase 1		Phase 2		Phase 3	
Joint	Direction	Mode	Direction	Mode	Direction	Mode
Hip	Extension	Active	Extension	Free	Extension	Active
Knee	Flexion	Free	Extension	Free	Extension	Active

Figure 6. Joint part detection and mode of dynamic motion in each Phase: (a) walking, (b) standing up and (c) going up stairs.

are performed reacting to the FRF. Therefore, it would be effective to determine Phase-shift timing from the FRF.

3.3.1. Walking. We focus on the condition of the left FRF for Phase-shift timing of the right leg. During walking the right leg swings (Phase 1 starts) when the rear part of the left foot contacts with the ground surface (Fig. 5a). At this time, we should be able to detect the FRF at the rear part of left foot and this detection can be used as an indication for the start of Phase 1. On the other hand, Phase 2 starts when the right foot contacts with the ground surface while the ground contact part of the left foot shifts to the front. As the result we should be able to detect the increasing in FRF at the front part of the left foot and use it as the indicator for the start of Phase 2. Based on these characteristics of the FRF, we set thresholds for FRF at the front and rear part of the left foot (V_{lf}, V_{lr}) which indicate the ground contact. If the FRF value at the front part of the left foot (f_{lf}) exceeds the threshold of the FRF at the front part of the left foot (V_{lf}), Phase 1 shifts to Phase 2. Subsequently, if the FRF value at the rear part of the left foot (f_{lr}) exceeds the threshold of the FRF at the rear part of the left foot (V_{lr}), Phase 2 shifts to Phase 1. We use the same method to determine the Phase-shift timing of the left leg based on the FRF of the right foot. The flow chart of the Phase Sequence in walking power assist for the right leg is shown in Fig. 7a.

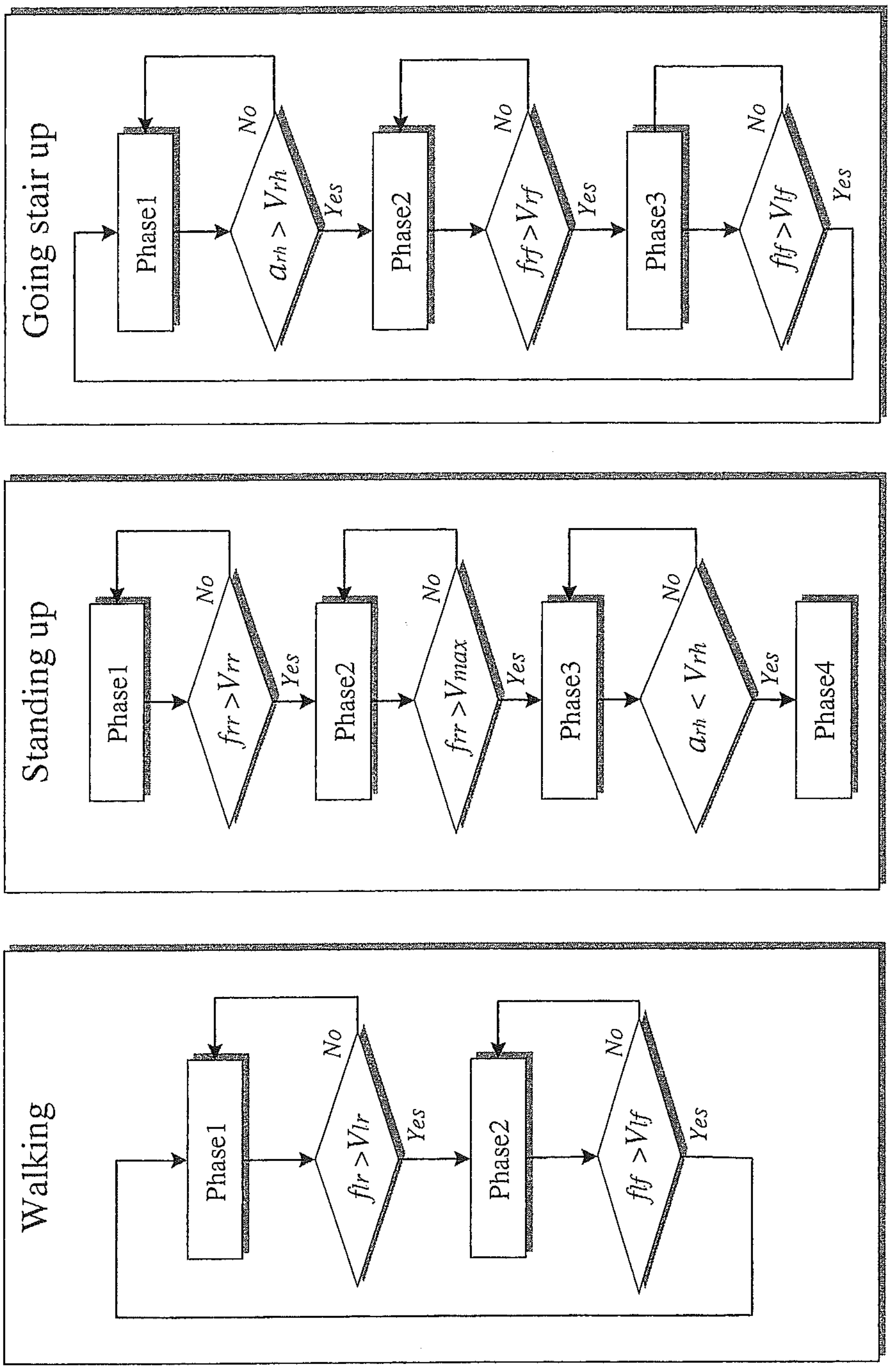


Figure 7. Flow charts for Phase Sequence: (a) walking, (b) standing up and (c) going up stairs.

3.3.2. Standing up. We use the condition of the FRF at the rear part of the right foot (f_{rr}) and the right hip joint (a_{rh}) in order to determine Phase-shift timing for standing up motion. As shown in Fig. 5b, when the upper body is slightly bent forward from the sitting position (Phase 2 starts), f_{rr} is detected. At this time, this detection can be used as an indication for the start of Phase 2. Phase 3 starts when f_{rr} exceeds approximately 80% of the maximum value. The hip and knee joints are extended in Phase 3. Finally, the standing up motion settles down in Phase 4 as the standing position. Based on these characteristics of FRF, we set thresholds for FRF at the rear of the right foot which indicate the condition for the shifts to Phase 2 and Phase 3, respectively. If f_{rr} exceeds the threshold (V_{rr}), Phase 1 shifts to Phase 2. Subsequently, if f_{rr} attains 80% of the maximum value (V_{max}) which is obtained from previous data of motion without the power assistance, Phase 2 shifts to Phase 3. Finally, if the right hip joint angle (a_{rh}) attains the predefined threshold (V_{rh}), Phase 3 is changed into Phase 4. The flow chart of the Phase Sequence in standing up power assist is shown in Fig. 7b.

3.3.3. Going up stairs. We use the condition of FRF at the front part of both feet and the right hip joint for the determination of Phase-shift timing of the right leg. As shown in Fig. 5c, during the going up stairs motion, the right leg is lifted up (Phase 1 starts) when the front part of the left foot contacts with the stair surface. At this time, we should be able to detect the FRF at the front part of the left foot and this detection can be used as an indication for the start of Phase 1. Phase 2 starts when the right hip joint angle attains the maximum value. Phase 3 starts when the right foot contacts with the ground surface. As a result we should be able to detect FRF at the front part of the right foot and use it as the indicator for the start of Phase 3. Based on these characteristics of the FRF and joint angle, we set thresholds for the FRF at the front part of the left and right foot (V_{lf}, V_{rf}) and the right hip joint angle (V_{rh}). If the FRF value at the front part of the left foot (f_{lf}) exceeds the predefined threshold (V_{lf}), Phase 1 starts. Subsequently, if the hip joint angle value (a_{rh}) attains the threshold (V_{rh}), Phase 1 shifts to Phase 2. If the FRF value at the front part of the right foot (f_{rf}) exceeds the predefined threshold (V_{rf}), Phase 2 shifts to Phase 3. We use the same method to determine the Phase-shift timing of the left leg based on the FRF at front part of both feet and the left hip angle. The flow chart of the Phase Sequence in going up stairs power assist is shown in Fig. 7c.

4. EXPERIMENTAL VERIFICATION

4.1. Method

The power assist experiments described in this section were performed with the Phase Sequence control defined in Section 3. In these experiments, power assist was performed for the active mode in each Phase by using HAL-3. The assist torque in each active mode was adjusted according to each Phase on the basis of the wearer's feedback suggestions. The assist torque pattern of each active mode is generated as a rectangular wave. Each threshold of the FRF for the Phase shift is determined based on the exoskeleton performance without any power assist. On the other hand, we set the threshold angles depending on the height of the chair or the stair used in this experiment. There is certain range for the thresholds necessary to transit Phases successfully. We determined the thresholds so that they could fall within this range. If the height of the chair or the stair was slightly different, Phases were successfully transited by using the thresholds.

It is considered that the assisted muscle's activation level during adequate power assist motion is reduced compared to the activation level of the motion without the power assist. However, the muscle's activation level during power assist motion with a time-delay of Phase switching is increased compared to that of the motion without the power assit. The muscle's activation level during each Phase is defined using the average myoelectric signal:

$$A = \frac{1}{T_\mathrm{p}} \int_{t_\mathrm{s}}^{t_\mathrm{s}+T_\mathrm{p}} E(t)\,\mathrm{d}t, \tag{1}$$

where $E(t)$ is the calibrated myoelectric signal and is a quatitative value related to the joint torque generated during the isometric contraction of the muscle; t_s and T_p represent the start time and the period of each Phase. The effect of the power assist was evaluated by comparison between the muscle's activation level with and without the power assist. The wearer was a normal 28-year-old male. Assist torque values were practically measured from the electric current consumed by the drive motors.

4.2. Experimental results

4.2.1. Walking. Thresholds of the FRF at the front and rear part of each foot for Phase shift were set as 640 and 180 N, respectively. The assist torques of the hip joint on the active modes on Phases 1 and 2 were 8 N, respectively. Figure 8a shows the joint angles, myoelectric signals and assist torques for the hip and knee joint, and the FRF in the front and rear parts of both feet during walking with power assist. The relation between the hip joint angle and the assist torque indicates smooth Phase shift. Figure 9a shows the averages of the muscle's activation levels at the hip flexor in Phase 1, and the hip extensor in Phase 2, during 10 steps of the right leg with and without power assist, respectively. It is obvious that the muscle's activation levels with the power assist are reduced compare to those without power assist.

4.2.2. Standing up. Thresholds for shifting to Phases 2, 3 and 4 were set as 60 N, 350 N and 0.1 rad, respectively. The assist torques of the hip and knee joints on the active modes of Phases 1 and 2 were 24 N, respectively. Figure 8b shows the joint angles, myoelectric signals and assist torques for the hip and knee joint, and the FRF at the rear parts of the right foot during standing up with power assist. The

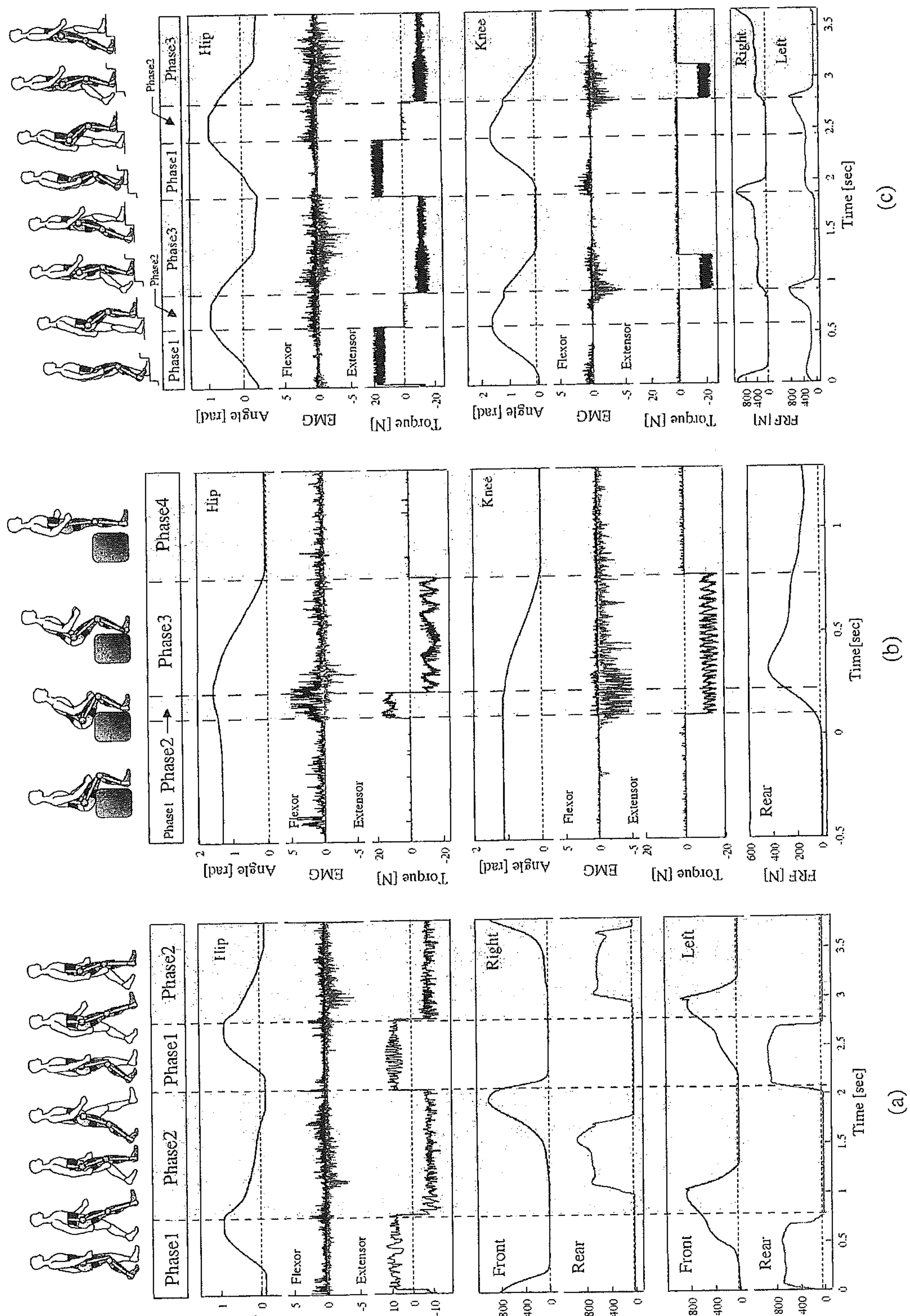
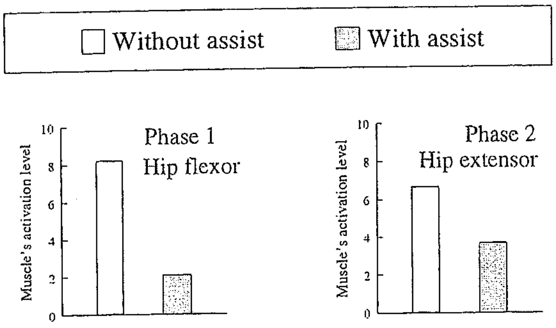
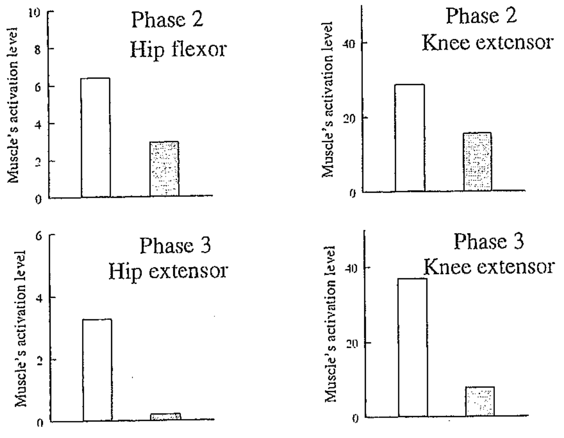


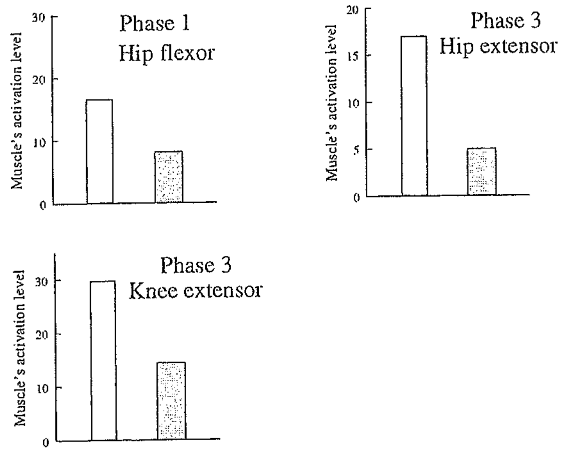
Figure 8. Exoskeletal performance: (a) walking, (b) standing up and (c) going up stairs.



(a) Waking



(b) Standing up



(c) Going up stairs

Figure 9. Comparison of the muscle activation levels for each Phase with and without the power assist.

behavior of the hip and knee joint angles is similar to that without the power assist (Fig. 5b). Figure 9b shows the averages of the muscle's activation levels at the hip flexor and the knee extensor in Phase 2, and the hip and knee extensor in Phase 3, during standing up with and without power assist 10 times, respectively. It is clear that the muscle's activation levels with the power assist are reduced relative to those without power assist.

4.2.3. Going up stairs. Thresholds for shifting to Phases 1, 2 and 3 were set as 160 N, 0.9 rad and 160 N, respectively. The assist torques of the hip joint on the active modes of Phase 1 were 16 N. Those of the hip and knee joints on Phase 3 were 12 and 20 N, respectively. Figure 8c shows the joint angles, myoelectric signals and assist torques for the hip and knee joint, and the FRF at the front part of both feet during going up stairs with power assist. It is clear that each Phase is shifted by using the predefined thresholds for the FRF at the front part of both of the feet and the hip angle. Figure 9c shows the averages and standard deviations of the muscle's activation levels at the hip flexor in Phase 1, and the hip and knee extensor in Phase 3, during 10 steps of the right leg with and without power assist, respectively. It is clear that the muscle's activation levels with the power assist are reduced relative to those without power assist.

4.3. Discussion

For each task, the hip and knee angle joints with power assist as shown in Fig. 8 were similar to those of motions without power assist as shown in Fig. 5. It is obvious that the Phases for each task were successfully transited and the tasks were realized. The activation levels of the muscles assisted in active mode were significantly reduced relative to the activation levels without the assistance (Fig. 9). This fact indicates that the autonomous motions generated by an exoskeleton robot effectively move the human's leg without any time-delay in Phase swiching. Therefore, we confirmed that the Phase Sequence method based on the modes of muscle activity is effective as an assist method by using autonomous motion of the exoskeleton robot HAL-3.

In the case of walking, we performed the assist including the active mode for the hip joint necessary to move forward and confirmed the feasibility. The assist including the passive and free mode for the knee joint was not performed. Further research should focus on adapting the power assist for the passive mode and the free mode by using the impedance control method [12].

The assist torque in each Phase was determined based on feedback suggestions from the wearer. We estimated the effectiveness of the assist by using the muscle's activation level. The reduction of the muscle's activation level differed according to the Phases. We have developed a method to adjust the assist torque on the basis of the muscle's activation level as a controlled variable [13]. The controller adjusting the assist torque in each Phase will provide more comfortable power assist for the wearer. Moreover, it is important to secure the safety of users of HAL. We have to note the risk of falling over caused by tripping, etc. We have studied the autonomous posture control method to prevent falling in parallel with the power assist methods.

In the near future, we will develop an autonomous motion assist method that applies the posture controller. Considering individual variations, we will use this method depending on the users of the HAL. In the case of a normal user, the thresholds for Phase-shift timing are determined based on the motion analysis data of individual users of the HAL. It is not necessary to measure the basic EMG signal because the FRF value and joint angle are used as indications for Phase-shift timing. The thresholds for the Phase-shift timing are determined according to individual users. However, it is convenient to use the EMG signal in order to analyze the muscle's condition during different motions or estimate the effectiveness of the assist. In the case of a handicapped user, the motion assist should be performed according to the standard motion. The standard motion has to be statistically determined based on motion analysis data of a number of normal persons. We plan to report on this issue in the near future.

5. CONCLUSIONS

In this research we proposed an assist method on the basis of autonomous motion of an exoskeleton robot. In order to perform tasks (walking, standing up and going up stairs) autonomically, we used Phase Sequence control, which generates a task by transiting some simple basic motions called Phases. Tasks were divided into some Phases during the tasks performed by a normal person. The joint moving modes were categorized into active, passive and free modes based on the characteristic of the muscle force conditions. The autonomous motion which HAL generates in each Phase were designed according to one of the categorized modes. FRFs and joint angles were adopted to transit each Phase. Power assist experiments were performed by using autonomous motion with a focus on the active mode. The experimental results showed that the muscle activation levels in each Phase were reduced. With this, we confirmed the effectiveness of the proposed assist method.

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Full paper

Estimation of a physiologic strategy based on a mathematical model for assisting and substituting cardiac functions by a robotic artificial heart

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Abstract—In order to assist and substitute diseased cardiac functions, it is important to develop a robotic artificial heart capable of adapting to physiologic changes in the same way the natural heart does. In this paper, we propose a method of estimating a physiologic strategy based on the systemic circulation model in order enable the effective medical care and therapeutic control using an artificial heart. The physiologic strategy is estimated by the internal physiologic parameters identified from a mathematical model using online parameter identification with the δ operator and the resonant frequency is calculated from the identified physiologic parameters. In the computer simulation study, the physiologic parameters and the resonant frequency could be identified precisely. During the animal experiment, the estimated physiologic parameters and resonant frequency corresponded to the physiologic changes. In the terminal stage of the animal experiment, the physiologic parameters were changed — mean peripheral resistance was increased from 0.4 to 1.0 mmHg s/ml and mean aortic compliance was decreased from 0.50 to 0.20 mmHg/ml. The resonant frequency was synchronized with the heart rate. As a result, we estimated the physiologic strategy as the changes in the identified physiologic parameters online to assist and substitute diseased cardiac function by a robotic artificial heart.

Keywords: Artificial heart; systemic circulation model; parameter identification; resonant frequency; δ operator.

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1. INTRODUCTION

The left ventricular assist device (LVAD) and total artificial heart (TAH) have been developed for the purpose of assisting and substituting diseased human cardiac function, respectively [1–4]. The inlet of the LVAD is connected to the apex of the left ventricle and the outlet of the LVAD is connected to the aorta. By delivering the blood flow from the left ventricle to the aorta, it assists diseased cardiac function of the left ventricle. TAH replaces the human heart and substitutes the diseased cardiac functions. However, a control method for an artificial heart, like that for the human heart, has not been defined clearly [5]. The operators have had to remain in the intensive care unit to observe the condition of the artificial heart and the patient, and manual intervention is required for controlling the artificial heart.

In our laboratory, we have developed a robotic artificial heart with non-invasive sensors [6, 7] and a network monitoring system [8] in order to use the artificial heart not only in medical care at a hospital, but also in home health care at the patient's home. The outline of this system is shown in Fig. 1. Using the non-invasive sensors, the robotic artificial heart could measure aortic pressure (Aop), blood flow, hematocrit, oxygen saturation and temperature non-invasively. These measured data were collected by a data acquisition server. Using the remote monitoring system, the robotic artificial heart could provide a remote monitoring website to enable medical staff to manage the condition of the patient and the artificial heart anytime from

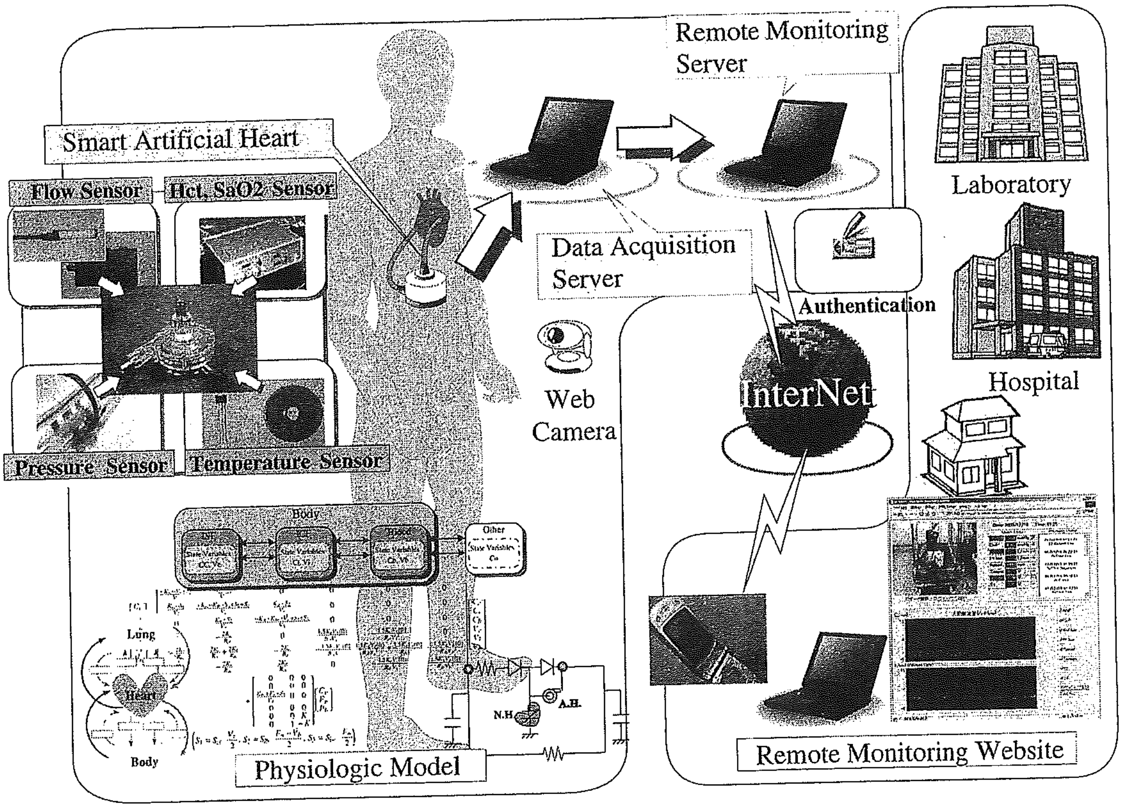

Figure 1. Outline of the robotic artificial heart with a smart artificial heart and a remote monitoring system capable of being used not only in the hospital, but also in the patients' home.