Stimulus adjustment protocol for FES-induced standing in paraplegia using percutaneous intramuscular electrodes

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Communications

Stimulus Adjustment Protocol for FES-Induced Standing in Paraplegia Using Percutaneous Intramuscular Electrodes

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Abstract—**The desirable upright posture for standing via functional electrical stimulation (FES) is defined based on simulation results using a link model in the sagittal plane. The criterion for the posture selection is the minimization of the sum of the squared joint flexion moments caused by gravity. The stimulus intensities are adjusted systematically to attain the defined upright posture. Controlled standing was achieved in a Th7 and a Th8-level spinal-cord-injured paraplegic individuals without joint contracture, by using the stimulus adjustment protocol. A practical standing without any bracing devices was obtained, with the vertical upper extremity support of less than four percent of the body weight, and with single-hand-support attainable. The maximal durations of standing were 30 minutes in both cases.**

I. Introduction

With conventional technology, the mechanical brace has been the only system to provide functional standing for individuals with paraplegia. Functional electrical stimulation (FES) offers another choice in achieving such standing [1]–[3]. The feasibility of standing and walking via FES methods has been demonstrated using surface and percutaneous electrodes, with or without bracing. Hybrid systems combining bracing with electrical stimulation have also produced stable standing and walking functions [4], [5]. The brace has played a significant role in increasing postural stability and reducing muscle fatigue in these studies.

On the other hand, the development of percutaneous intramuscular electrodes and the corresponding implantation techniques and stimulator design have provided selective stimulation of a number of muscles [6]–[8]. Some studies have attempted to restore upper and lower extremity functions based on this percutaneous electrode technology. Selective activation of muscles via a number of electrodes has made it possible to attain higher-order functions such as grasping or stairway climbing [9]–[11]. These references have described the electrodes, the technique for the electrodes implantation, and other engineering and medical concerns. Since the stimulus intensities for the separate electrodes were adjusted experimentally, a long time was required to improve the restored FES motion, even though a standard stimulation pattern was used that had been obtained from EMG analysis of the movement of neurologically intact individuals. This extensive time requirement increased the mental and physical load on the patient as well as the clinical cost. So the problem to be solved was to establish adjustment protocol that could select appropriate stimulus intensities for the percutaneous electrodes in reasonable time.

The purpose of this study is the establishment of an efficient stimulus adjustment protocol based on 1) definition of the desirable upright posture for FES-induced standing by a link model simulation, 2) translation of the defined posture into the clinical applicable stimulus adjustment protocol to attain the posture via FES, and 3) evaluation of the defined protocol through clinical application.

II. Problem Description

The practicality of standing via FES has been limited by postural instability and a short standing duration. While the "C" shaped posture with hyperextension of hip joint is a reasonable posture to obtain hip joint stability [12]. As shown in Fig. 1(a), the posture with excessive hyperextension of the hip joint is frequently observed in the standing with knee-ankle-foot-orthosis. This posterior displacement of the center of gravity by hip hyperextension brings a large knee flexion moment. In contrast, an upright posture with hip flexion, as shown in Fig. 1(b), can be often observed in FES-induced standing with the stimulation of the quadriceps femoris. As shown in Fig. 1(c), the center of gravity is located in front of the ankle joint in upright posture of an able-bodied individual. The first problem is the selection of the standing mode among these three standing modes. The posture should be suitable for FES-induced standing using percutaneous electrodes without a bracing device, from the viewpoint of postural stability and standing endurance. In this study the upright posture that minimizes the required joint moment was selected as the solution for this problem by using a model simulation method. Since the joint moment caused by gravity should be compensated by the muscle contractile force, the less requirement of joint moment is more desirable.

In conventional FES-induced standing of paraplegic individuals, two pairs of surface electrodes have been used to stimulate the quadriceps femoris. The four muscles of the quadriceps femoris are stimulated simultaneously with this method. In contrast, the stimulators for the percutaneous electrodes can activate up to 48 different channels [11]. A 30 channel system has been developed and it is commercially available in Japan [10]. By using this multichannel stimulator system, it is possible to stimulate the four muscles of the quadriceps femoris separately. This method means that the stimulus intensities of the rectus femoris, which has hip flexion effect, and the other three muscles, which have no effect on hip joint, can be adjusted selectively. However, the question of how to adjust the stimulus intensities of these electrodes must be resolved. In addition, the balance of the contractile forces of the antagonistic muscles, such as the rectus femoris and the gluteus maximus, need to be adjusted adequately to attain a suitable posture. While our stimulation system provides a sufficient number of stimulation channels for antagonistic muscle control, the method to adjust the balance of the stimulation channels has not been discussed. The second problem is the establishment of the stimulus adjustment protocol, with reasonable time and procedure, to attain the desirable upright posture. This paper describes a stimulus adjustment protocol that addresses these questions.

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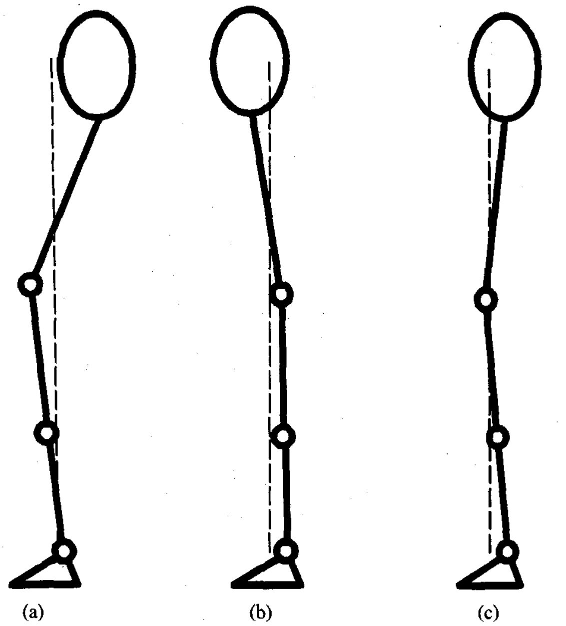


Fig. 1. Three standing modes classified according to the order of horizontal positions of knee, hip and center of gravity. (a) is the strongly hip-hyperextended posture as observed in standing with knee-ankle-foot-orthosis, (b) is the hip-flexed posture as observed at the beginning of FES-induced standing. (c) is the upright posture of able-bodied individual.

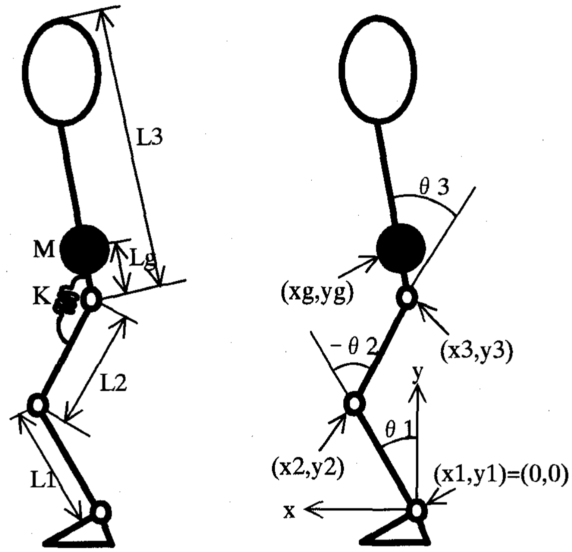


Fig. 2. A three-link model with a point mass for the simulation study. The ligaments around the hip joint were modeled as a monodirectional elastic element. The range of the knee joint angle was limited for not to be hyperextended. The foot was assumed to be fixed on the floor.

III. Methods

A. Standing Mode Selection by Model Simulation

A static simulation study in the sagittal plane was carried out using the three link model shown in Fig. 2 in order to determine the desirable standing mode for FES-induced standing among three standing modes shown in Fig. 1. The length of each segment was determined by the measurement of a skeletal specimen. A point mass was assumed to be located at 57% of the height. From the viewpoint that the requirement for upper extremity in supporting body weight is undesirable, the constant support force given by upper extremities was assumed to be zero. In other words, upper extremities were considered to be used only for the compensation of the postural sway.

The following three conditions were added to the model based on the human skeletal structure. 1) The sagittal position of the center of gravity should be above the center of the foot. This condition was measured in normal standing using a force platform. This position gives a margin against the anterior–posterior fluctuation. 2) The moment in the direction of the knee joint hyperextension was neglected, because the moment which affects this direction is compensated by the skeletal structure and the ligaments around the knee joint. Therefore, contraction of the quadriceps is not required in this posture. 3) The ligaments around hip joint were modeled as a nonlinear elastic element which affects only against hip joint hyperextension, because the ligaments generate passive force against hip joint hyperextension. The criteria for determining a suitable posture was the minimization of the sum of the squared moment of ankle, knee and hip joints caused by gravity. The calculation procedure of the simulation is described follows.

The ankle joint moment m_1 is given as the product of the sagittal position and the mass.

$$m_1 = x_g\ M. \tag{1}$$

Here, x_g is given by the previously mentioned condition 1). Then, m_1 takes constant value. The knee joint moment can also be described as follows while x_2 means the sagittal knee joint position.

$$m_{2g} = (x_g - x_2)\ M. \tag{2}$$

As described in condition 2), the knee joint must not be hyperextended.

$$\theta_2 \leq 0. \tag{3}$$

And, the knee joint moment was neglected while the knee joint moment acts for the knee joint hyperextension.

$$m_2 = \begin{cases} 0 & (\theta_2 = 0 \text{ and } x_g > x_2) \\ (x_g - x_2)\ M & (\text{other cases}) \end{cases}. \tag{4}$$

The hip joint moment by gravity m_{3g} is also given by the mass and the sagittal position of the hip joint and center of gravity.

$$m_{3g} = (x_g - x_3)\ M. \tag{5}$$

The hip joint moment by the ligament m_{3k} is also given when the hip joint is hyperextended as mentioned in condition 3).

$$m_{3k} = \begin{cases} K\theta_3 & (\theta_3 < 0) \\ 0 & (\theta_3 > 0) \end{cases}. \tag{6}$$

Therefore, the total hip joint moment is given as follows.

$$m_3 = \begin{cases} (x_g - x_3)\ M + K\theta_3 & (\theta_3 < 0) \\ (x_g - x_3)\ M & (\theta_3 > 0) \end{cases}. \tag{7}$$

The sum of the squared moments of the three joints Q is given as follows. The upright posture which gives minimal value of Q was selected.

$$Q = \sum_{i=1}^{3} mi^2. \tag{8}$$

The simulations were performed with several values of the hip ligament elasticity because the exact value is unknown. The parameters of the model is listed in Table I.

TABLE I
PARAMETERS OF THE LINK MODEL FOR THE SIMULATION STUDY

L1(m)	L2(m)	L3(m)	Lg(m)	M(Kg)	K	Xg(m)
0.33	0.38	0.72	0.11	40	$0\sim\infty$	0.05

B. Translation into Stimulus Adjustment Protocol

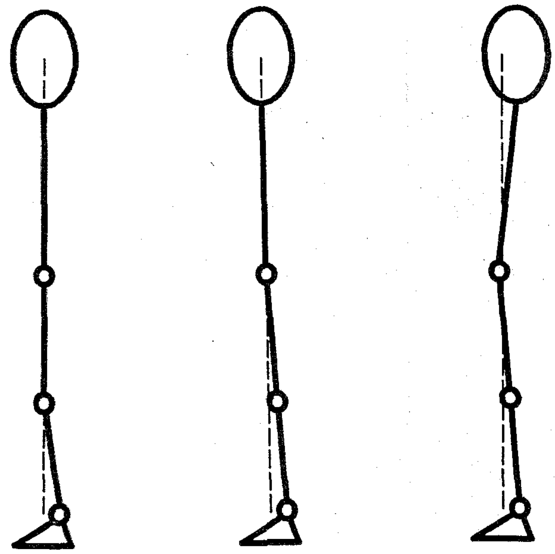
The able-bodied-like standing mode was selected as the desirable posture by the model simulation, as described in the following section. The upright posture of the model was interpreted into the stimulus adjustment protocol in the sequence of the knee, hip, and ankle joints. The stimulation sites for standing were classified into five groups of 1) the femoral nerve trunk (all of the quadriceps), 2) the vastus medialis, lateralis, and intermedius, (nerve branch), 3) the gluteus maximus, 4) the common peroneal nerve trunk (the tibialis anterior and other muscles), and 5) the gluteus medius. In this study, the stimulus adjustment protocol for the four groups which concern sagittal motion, except for the gluteus medius, was discussed. The margin in the knee extension moment was taken for not causing knee collapse when larger knee flexion moment than assumed moment is applied by backward swaying. The index for setting was defined in the manual muscle test method [15], because it is more clinically practical to be applicable without any special measurement equipment. The joint moments gained by the stimulation were measured after the adjustment by the proposed protocol.

C. Paraplegic Standing

A commercially available portable stimulator (FESMATE 1000 CE1230, NEC, Tokyo) and percutaneous electrodes [9] were employed. The bipolar, monophasic and voltage-controled pulse train was used for stimulation. The stimulus parameters were 20 pulses/s in frequency, 200 μs in pulse duration. The amplitude was modulated up to 15 V. In this study, the open-loop control scheme was applied by the stimulation using previously set stimulus data [10]. This method can not change stimulus intensities in proportion to the joint angle or the muscle contractile force in contrast to a closed-loop control system [13], [14]. But this method requires no sensors and can be constructed with minimal hardware. Therefore, this open-loop system seems to be most clinically practical given by the conventional technology. The average numbers of the electrodes used for standing were 20. All thirty stimulation channels were used by the electrodes since we also implanted in other muscles for walking such as hamstrings.

D. Posture Evaluation

The upright postures of two individuals with paraplegia were evaluated by using a video motion analyzer (Quick-Mag, OKK, Tokyo). The upright postures of one minute quiet standings were measured and averaged during the standing with the support of upper extremities between parallel bars. The sample interval of the system is 33 ms and the resolution under the used camera position is 0.5 mm (512×512 pixels). The landmarks were positioned on the malleolus lateralis, the caput fibulae, the trochanter major, and the acromion. A ground reaction force measurement system was also used simultaneously. The system was constructed of a force platform (ECG-1010D, Kyowa dengyo, Tokyo) with 2.5 mm position error and 1 Kg force error in maximal, a computer with 12 b A/D converter board, and original software for data acquisition and analysis. The sample rate of the system is 10 ms. The joint moments by gravity were calculated from the measured values of the position and the ground reaction force using (1), (2), and (5).



(a) (b) (c)

Fig. 3. Solutions of simulation study as desirable postures. Both (a) "unstable" posture and (b) "knee-locked" posture are the solutions without the effect of hip joint stiffness. The effect of hip joint stiffness of 2500 Nm/rad changed "Knee-locked" posture to the posture with hip joint hyperextension as shown in Fig. (c).

The postures of five neurologically intact individuals were also measured to obtain a reference posture. The averaged posture was compared to the postures of the individuals with paraplegia.

IV. RESULTS

A. Selected Standing Mode

Fig. 3 shows the desirable upright postures under FES that was defined through the link model simulation. Both Fig. 3(a) and 3(b) are the solutions without the effect of the elasticity of the hip joint ligaments. Two solutions were given because of the nonlinearity of the model around the knee and hip joints. Fig. 3(a) is the "unstable" posture which has perpendicular thigh and trunk. Fig. 3(b) is the "knee-locked" posture. The hyperextension angle of the hip joint in "knee-locked" posture was increased as the hip joint elasticity increased to 3000 Nm/rad. The "unstable" posture was the same as Fig. 3(a). The "knee-locked" posture with the elasticity of 2500 Nm/rad are shown in Fig. 3(c) as an example. The stability of the hip joint in this posture is greater than it in the "unstable" posture, because of the antagonistic action of the gravitational force and the elastic force of the ligament. The order of the joint horizontal positions in this posture is the same as it is in the able-bodied posture of Fig. 1(c). Therefore, the standing mode with minimal joint moments was found to be the able-bodied standing mode.

B. Stimulus Adjustment Protocol

According to the selected posture in Fig. 3(c) and the able-bodied posture in Fig. 5 and Table IV, the contraction of the quadriceps is not required to compensate the knee flexion moment by gravity. The moment for hyperextension is compensated by the skeletal structure in these standings. The margin to prevent the knee collapse was taken against the backward sway. The assumed maximal backward displacement of the center of gravity was 35 mm. This assumption means that the center of gravity is located just above the ankle joint while in the maximal sway. In this posture, the knee flexion moment for one lower extremity is estimated as 10 Nm, that is the half of the product of the averaged weight 58 Kg and the displacement 35 mm. The clinical index was selected as "good" in manual muscle testing method. This is the grade 5 of 6 grades. The grade "good" means less force than "normal" and also more force than "fair," while "fair" grade gives complete extension against gravity. The additional problem around knee joint is the ratio of the stimulus intensities of the femoral nerve trunk and the nerve branch for the vastus muscle group. Both nerve trunk stimulation and nerve branch stimulation around the end plate are feasible with percutaneous electrodes. Stimulation at the femoral nerve trunk provides a strong knee extension because it activates all muscles of the quadriceps, but it has a hip flexion effect because the rectus femoris is activated. By comparison, selective stimulation of the vastus muscles at their nerve branches provides pure knee extension. Therefore, the femoral nerve trunk was stimulated only when the knee extension moment produced by the vastus muscles was insufficient.

The flexion moment of hip joint of the normal subject was 4.1 Nm in average (Table IV). This should be compensated by the ligaments. Another consideration is the hip flexion effect of the rectus femoris. This moment should be compensated to avoid the "hip-flexed" posture as seen in Fig. 1(b). The gluteus maximus was stimulated to compensate the flexion moment.

The gravity center was located in front of the ankle joint in able-bodied subjects. The requirement for the muscles around ankle joint is plantar-flexion of 17 Nm (Table IV). It means the contraction of the triceps surae. However, the triceps surae spasticity is frequently observed in paraplegic individuals. In such cases, excessive plantar-flexion moments can be generated beyond the requirement to maintain a desirable upright posture. Thus, the stimulation of the peroneal nerve to activate the tibialis anterior and the peroneus muscles would be required to compensate for the excessive plantar-flexion moment. In this study, even in the cases without the triceps spasticity, the triceps surae was not stimulated to avoid aggravating the spasticity. The index was selected as the rectangle of ankle joint while the knee joint is fully extended. The ankle joint will be dorsi-flexed 6° from the set position when the individual takes the defined posture that is the same as it of able-bodied subjects. The passive plantar-flexion moment is expected to be given by this slight dorsi-flexion. The error between the obtained ankle dorsi-flexion moment and the exact required moment, 35 Nm plantar-flexion (average of able-bodied standing in Table IV), can be compensated by the upper extremities.

The flow of the adjustment protocol of the stimulus intensities for one lower extremity is shown in Fig. 4. According to this flow chart, the stimulus intensities were manually adjusted by an operator. The muscle contractile forces were evaluated in manual muscle testing method [15], while the individual with paraplegia was in the supine position. The adjusted amplitudes were stored in the floppy disk and the memory in the stimulator. The adjustment was basically once before standing up.

C. Adjusted Joint Moment of Stimulated Muscles

The profiles of two paraplegic individuals are summarized in Table II. Paraplegic individuals who have suffered from complete spinal cord injury in the middle thoracic level were selected in order to discuss the validity of the proposed stimulus adjustment protocol. The joint moment obtained by the muscle stimulation after the stimulus adjustment are shown in Table III. The joint positions, while the joint moment were measured, were not the same as the joint positions used in the protocol definition, because of the expedience for the moment measurement. For example, the isotonic extension moment of knee joint can not be measured using passive measurement device, such as load-cell or others, under the knee joint full extension. The obtained knee extension moment was greater than the desired value in the protocol, because of the joint position. The actual moment at the extended position will be less than the shown values. Note that this extension moment is the margin against the backward sway. The error between the adjustment index and the actual moment does not affect seriously, because of the knee locking mechanism. The difference between the hip flexion and extension moment are more effective. The joint moment caused by the difference must be compensated by the upper extremities to maintain the adequate posture. In the right leg of the case A, the undesirable hip flexion moment of 4.7 Nm were given by the stimulation of the rectus femoris. The compensation by the gluteus maximus stimulation was 17% excessive. The joint moment to be compensated by the upper extremities significantly decreased by the stimulation of the hip extensor. The hip abduction moments by the gluteus medius were also shown for a reference.

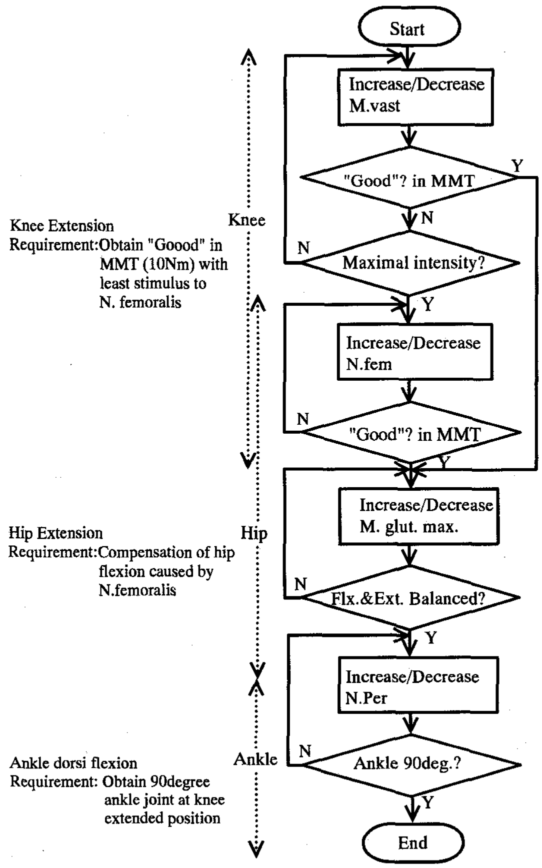


Fig. 4. The flow of the stimulus adjustment protocol per one leg, generated through the discussion in the sagittal plane. This protocol was applied while the subject was in supine position. The right and left gluteus medius were also stimulatied with constant and balanced stimulus intensity.

D. Standing of Paraplegic Individuals Via FES

The upright postures while standing between parallel bars using FES of these individuals are presented in Fig. 5. The order of the horizontal position of joints was ankle, knee, and hip, except case B, who has an hyperextensive knee joint. The order was the same in the standings of the able-bodied subjects. It was found that the posture of the paraplegic individuals were more upright by comparing Fig. 5(a), (b), and (c). The hip joint moments were slightly greater than the average of the moment in able-bodied subjects (Table IV),

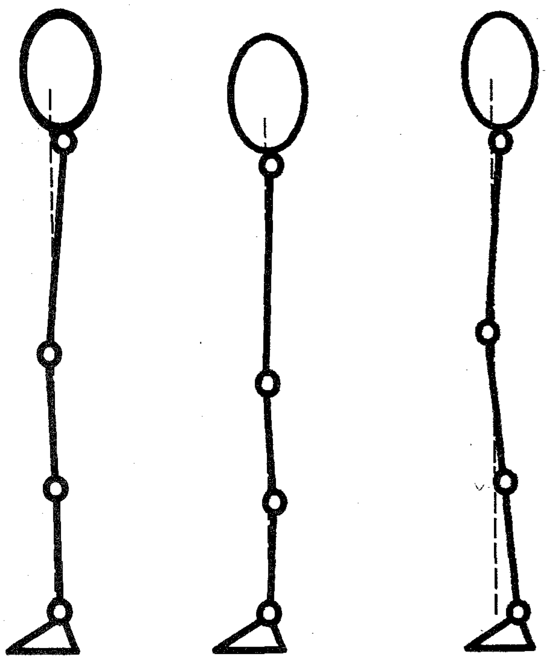


Fig. 5. The measured upright postures of FES-induced standings between parallel bars in (a) Th7, (b) Th8 level spinal-cord-injured paraplegic individuals and (c) the represent of the average joint angles obtained from the five able-bodied subjects, shown in Table IV. The length of the segments in (c) are the same as the length of the segments of the link model.

TABLE II
SUMMARY OF THE PATIENT PROFILES. BOTH CASE WERE WITH COMPLETE SPINAL-CORD-INJURED PARAPLEGIA

Subject	Sex	Level of Injury	Age (Yrs.)	Post Injury (Yrs.)	Time in Study (Mos.)	Standing Duration* (min.)
A	M	Th7	24	2	16	30
B	F	Th8	24	3	17	30

* Measured at 6 months after FES beginning.

but within the standard deviation. The knee joint moment in case B showed greater value than it in able-bodied because of the strong hyperextension. In contrast, the moment in case A showed less value because of the more perpendicular posture. In every case including able-bodied subjects, "knee-locked" posture was taken. The ankle joint moments indicate the most remarkable difference between the FES-induced standings and the able-bodied standing. The less dorsiflexion moment mean the backward offset of the center of gravity and more perpendicular posture as seen in Fig. 5.

The standing duration was thirty minutes, measured at six months after the beginning of FES usage. The standings with single arm support were also possible more than one minute in these two individuals as shown in Fig. 6. It was also possible to perform work with their free arm. The detachable cane system combined with a wheelchair can be seen.

V. DISCUSSIONS

The joint moment by gravity, except for the moments compensated by the skeletal structure or the ligaments, requires to be compensated by muscle contraction. The requirement for the large contractile force is obviously undesirable because it causes the property change by muscle fatigue and increases the energy consumption. The minimization of the sum of the squared joint moments was chosen as the optimization criterion in this study. The minimization of the joint moment is not equivalent to the minimization of the contractile force because of two reasons of the biarticular muscles and the coactivation.

TABLE III
JOINT MOMENTS OBTAINED BY THE STIMULATION. THE STIMULUS INTENSITIES WERE ADJUSTED BY THE PROPOSED PROTOCOL

Subject	Side	Knee Ext. (Nm)	Hip Flex. (Nm)	Hip Ext. (Nm)	Hip Abd. (Nm)	Ankle (deg)
A	R	19	4.7	5.5	2.4	0
	L	38	5.5	4.0	2.9	0
B	R	21	1.5	3.0	1.6	0
	L	16	1.5	1.5	1.6	0
Measured position		Flex. 60deg.	Flex. 10deg.	Flex. 10deg.	Abd. 10deg.	-

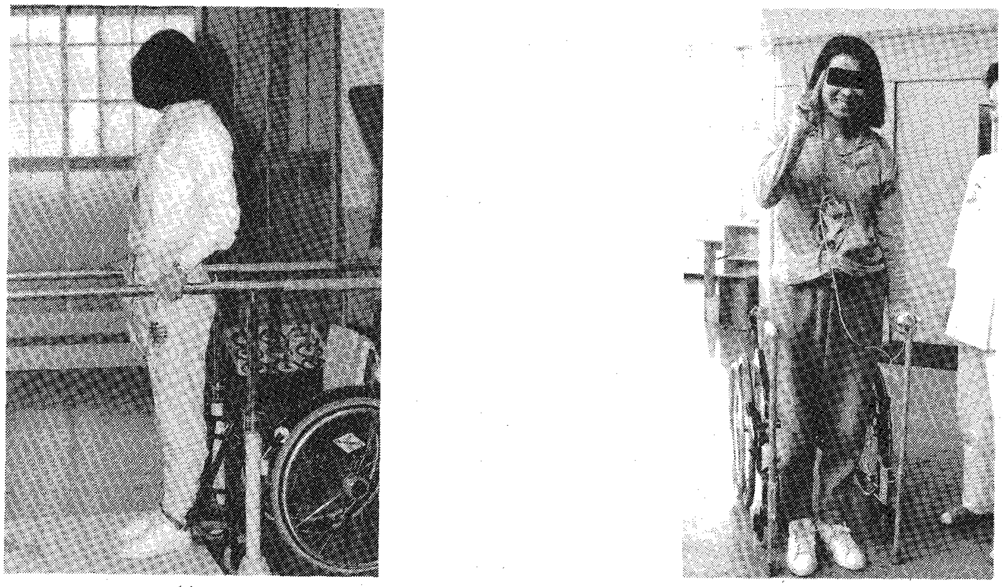
TABLE IV
UPRIGHT POSTURES OF TWO INDIVIDUALS WITH PARAPLEGIA (A, B) AND AN AVERAGE OF FIVE ABLE-BODIED SUBJECTS

	Moment			Angle			Weight	
Subject	Ankle (Nm)	Knee (Nm)	Hip (Nm)	Ankle (deg.)	Knee (deg.)	Hip (deg.)	GRF (Kg)	Weight (Kg)
A	-6.9	-1.4	-7.4	3.7	2.2	-16.6	51.2	52.9
B	-0.3	-14.7	-5.9	-10.6	-24.4	-20.2	54.3	56.5
Average of able-bodied	-17.2 ±4.3	-8.7 ±3.1	-4.1 ±3.9	5.8 ±2.9	-0.8 ±2.7	-11.1 ±3.8	60.7 ±9.7	60.7 ±9.7

Positive values mean dorsal flexion in ankle joint and flexion in knee and hip joints. Value of moment is for one leg.

Biarticular muscle makes the relation complex between muscle force and joint moment. Biarticular muscle improves the efficience of the muscle force in lots of situations [20]. In respect of the standing, the hamstrings, the gastrocnemius, and the other biarticular muscles of the lower extremities have profitable effect on one joint but also have harmful effect on another joint. The rectus femoris has a feasibility to transfer the hip hyperextension moment into the knee extension moment, however it is empirically known that the stimulation of the rectus femoris frequently causes the undesirable hip joint flexion as shown in Fig. 1(b). Therefore, the stimulation of the biarticular muscles were basically avoided and the monoarticular muscles were mainly stimulated as far as possible.

In addition, if the counteraction between the flexion and extension moment does not occur (if antagonistic muscle pair does not contract simultaneously), the minimal joint moment is equivalent to the minimal muscle force. In the proposed FES-induced standing, the rectus femoris and the gluteus maximus, as hip flexor and extensor, stimulated simultaneously. The simultaneous contraction of the antagonistic muscle pair increases the joint stiffness. In the presented cases, the coactivation of hip flexor and extensor appears to affected the hip joint for stabilization. The coactivation level of the hip flexion and the extension moments which gave successful standing are shown in Table III. However, the optimal activation levels of antagonistic muscles for FES standing are unknown. When the desirable joint stiffness is given in addition to the desirable joint moment, the problem of antagonistic two muscles can be solved. The standard is expected to be given in the analysis of the posture of the able-bodied subjects. The upright posture with minimal muscle force requirement will be directly solved by using the standard value.



(a) (b)

Fig. 6. The side (a) and the frontal (b) views of the examples of the restored standing via FES in case B. The canes attached to the wheelchair can be seen in the right picture.

The knee extension moment that was adjusted by the proposed protocol was higher than the assumed margin (Table III). However, the quadriceps contraction was not required in the reconstructed standings of the paraplegic individuals (Table IV). The no requirement is supported by the fact that the thirty minutes standing was possible. It should be impossible to maintain upright posture for thirty minutes, if the muscle contractile force was used against the gravity. The muscle contractile force appears to be used just against the postural fluctuation. The continuous stimulation was applied as safety margin. The undesirable effect of continuous stimulation such as muscle fatigue can be avoided by reducing the total amount of stimulus through the use of a feedback control scheme such as an artificial reflex [4], [17].

The backward-leaning of the trunk was larger than it in the able-bodied subjects, as is well known [12]. However, the center of gravity was located in front of the knee joint in both cases. It is reported that the position of the center of gravity was behind the knee joint, in the study using a surface electrode stimulation of the quadriceps [16]. It seems that the trunk can be brought to a more upright position without instability by increasing hip joint stiffness through antagonistic coactivation. It can be said that the increase of hip joint stiffness is effective in attaining the able-bodied-like standing.

The more upright postures in the cases with paraplegia are confirmed by Fig. 5 and the ankle joint moments in Table IV. The upright posture might be attribute to the insufficience of the plantar-flexion moment because of the lack of the stimulation to the triceps surae. The enhancement of the triceps spasticity must be avoided. If the utilize of a brace does not conflict the requirement of the users, the hybrid system with ankle-foot orthosis [4] will be helpful to obtain the defined plantar-flexion moment at the defined ankle joint angle without the undesirable stimulation. It will offer the more forward-leaning posture like as able-bodied subject.

The average of the upright postures of the able-bodied subjects was utilized as a reference posture. The utilize bases on the assumption that the difference of the musculo-skeletal system is negligible between the kinematic property of paraplegic subjects and it of able-bodied subjects. The assumption is satisfied in except for the cases with joint contracture and other complications, and limits the candidates of the proposed protocol. The criteria for patient selection for the application of FES were discussed in a number of studies [1], [13], [16]. The essential criteria such as absence of peripheral nerve damage are basically identical among these studies. The candidate of the proposed protocol is more limited. The method to extend this protocol should be studied. However, the therapeutic effect of the electrical stimulation, such as the spasticity reduction and the muscle restrengthening, can be significant [18], [19]. For example, the range of motion in the ankle joint can be improved by the reduction of the spasticity in the triceps surae. It is at least required to remove the obstacles by the conventional rehabilitation treatment before the beginning of FES.

Most activities of daily living require both upper extremities, and two-hands-free standing is desirable. A feedback control scheme for the disturbance compensation will be necessary to attain such standing [21], [22]. Only the lower extremity muscles were controlled in most of the previous studies. The large hyperextension of hip joint that was observed in the postures of the paraplegic individuals, includes the posterior flexion of the spine. Therefore, trunk control will be required to compensate the spine flexion. The trunk control is indispensable even in the environment where the pelvis can be supported, such as kitchen sink. However, the control problem of the trunk is complicated because it has multiple degrees of freedom. Further study is required.

Clinical practicality of the restored standing is limited by the requirement for the environment. In the two individuals in this study, single hand support was sufficient to maintain the upright posture, however, double hand support was required to stand up and sit down. The detachable cane system still has a problem during the transportation of the wheelchair. The development of the more convenient support device for the transport is desired. An implanted stimulator is also an improvement for practicality, because it removes the daily disinfection of the body-entry point of the electrodes.

VI. Conclusion

The desirable standing mode for FES standing was selected as the posture of able-bodied subjects by using the link model simulation method. The simple stimulus adjustment protocol for the paraplegic

FES-induced standing with the percutaneous intramuscular electrodes was proposed. Thirty minutes of standing between parallel bars as well as the single hand support standing were attained without bracing. The optimization of the joint stiffness is necessary to directory determine the muscle force. The method to overcome the joint contracture should also be studied in order to extend the candidate of the protocol.

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A Portable Measurement System for Prosthetic Triaxial Force Transducers

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***Abstract*—A portable system was developed to process and store normal and shear stress data measured at thirteen sites at the stump-socket interface of lower-limb amputees. Forces and moments measured in the prosthetic shank were also processed and stored. Custom-designed sensors and small-size signal conditioning units described in detail elsewhere [1], [2] were mounted on the prosthesis to take the measurements. The data collection unit, described in detail in this communication, was a Motorola HC16 microcontroller with appropriate peripherals and custom-written software. A 4 Mbyte PCMCIA card was used for data storage. The system collected data for approximately 5 minutes of continuous monitoring or for segmented trials of durations specified by the user. Evaluation tests showed good linearity with minimal hysteresis, though there was some 10–90 mV peak-to-peak noise on the stored data. The principal source of noise was interference between wires in the belt pack instrumentation. The noise can be reduced in the future by changing from a wire-wrapped board configuration to a printed circuit design. Use of the system on an amputee subject showed data of similar magnitude to that collected with a stationary system (cables to a computer data acquisition unit). The system will be used to collect data to improve understanding of how prosthesis design features effect interface stress distributions and also to evaluate stump-socket finite element models. Finite element models are computer-based tools that potentially will predict interface stresses for proposed socket designs and thus enhance the prosthetic design process.**

I. INTRODUCTION

Interface stresses are a crucial aspect of prosthetic fitting. Prostheses must be designed to provide adequate support and stability for weight-bearing, but at the same time do not traumatize residual limb soft tissues. Overstressing of tissues leads to breakdown, a condition that restricts prosthesis use and thus worsens the disability. An important challenge in prosthetics is to design artificial limbs that induce acceptable interface stress distributions and avoid breakdown.

Interface normal and shear stress measurements have important uses in prosthetics. They help to improve understanding of how prosthesis design features effect the interface stress distributions [3], [4]. They also provide data for comparative evaluation of

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