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Effects of Stimulation Conditions and Waveforms on Muscle Contractile Characteristics

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Abstract: This study was designed to apply the stimulation system developed in our laboratory to investigate how the stimulation conditions affect the muscle contractile characteristics in the isometric condition as well as during the FES standing/walking. Four paraplegic and ten healthy subjects participated in this study, and their knee extensors were voluntary contracted or electrically stimulated to measure the muscle force and the fatigue index for different waveforms of the pulse train. We also investigated different combinations of the electrode positions during standing/walking.
It was confirmed that continuous and high-frequency stimulation causes faster fatigue than intermittent and low-frequency stimulation. Fatigue resistance was higher around the optimal muscle length than at a stretched position in healthy subjects, whereas the opposite was observed in paralyzed subjects. The paired *t*-test results with the level of significance at 0.01 indicated that the sinusoidal waveform generated the largest torque among the four typical waveforms. Although statistically not very significant, the sinusoidal waveform also generated, in general, the highest fatigue resistance at an intensity level below the supramaximal stimulation. One of the paraplegic subject who participated in the standing/walking program can now stand up for 1 minute and 50 seconds with the knee extensors, and walk for about 5 minutes at the speed of 12m/sec.

Key words: Functional electrical stimulation, Pulse waveform, Stimulation condition, Electrode position

INTRODUCTION

The first practical application of functional electrical stimulation (FES) to mobility recovery of the plegic was reported by Liberson et al. [1]. They used surface electrodes to electrically stimulate the peroneal nerve (PN) of fossa poplitea, induced dorsiflexion during the swing phase, and consequently avoided drop foot of hemiplegic patients. So far, significant time and effort has been devoted to paraplegic standing and walking [2-5], quadriplegic grasping and wrist motion [6-8], and so forth, in the area.

Standing and walking by means of FES has many advantages over conventional orthotic devices [9,10]. First, the user uses his/her own muscles, joints, bones, metabolic energy, and metabolic energy. In addition, standing up and moving the body can prevent atrophy, contracture, immobilization of joints, demineralization of bones, etc. and increase blood circulation and muscle force. Pressure sore can be also prevented by distributing the pressure acting on the body surface. In spite of all such advantages, many problems should be solved before FES can be easily applied to the patients' daily lives.

One of the major problems impeding easy application of FES is insufficient muscle force and muscle fatigue induced by electrical stimulation [11,12], defined as a force reduction in this study. Paralyzed muscles show disuse atrophy with time that decreases muscle force and causes muscles to fatigue easily. Electrical stimulation recruits muscle fibers in an order reverse from normal; when a muscle is electrically stimulated, fast fibers are recruited before slow fibers showing strong fatigue resistance, so that stimulated muscles fatigue faster than those under voluntary contraction [13-15]. Furthermore, slow muscles, i.e., muscle having more slow fibers than fast fibers, indicated almost the same mechanical characteristics as fast muscles a few weeks after spinal cord injury (SCI) [16,17]. Most researches agreed that high-frequency stimulation showed rapid fatigue due to failure of electrical propagation, whereas long-term low-frequency stimulation can prevent atrophy and reduce muscle fatigue [18-20].

In this study, we investigated contractile

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characteristics and fatigue characteristics of knee extensors of ten healthy subjects and four paraplegic patients under different stimulation conditions: continuous stimulation and intermittent stimulation, different stimulus parameters (pulse width, pulse amplitude, and frequency). We also investigated how the pulse waveform affected the muscle force and the fatigue characteristics. One of the paraplegic patients started an exercise program for FES walking described in [5], and participated in FES-induced standing and walking, and the results are briefly described in this paper.

METHOD

Subject Selection

Ten healthy subjects and four paraplegic patients participated in this study. The healthy subjects were selected among the undergraduate and graduate students in the authors' department, and they were decided to have no medical problems by physical and mental examinations. Their ages ranged from 18 to 25 with the mean of 20.7. The paraplegic patients were selected among the SCI patients in the Konkuk University Hospital, Choongju, and National Rehabilitation Hospital, Seoul, Republic of Korea. The rehabilitation medicine team did medical examinations, such as the bone density examination, computerized tomography of the lower extremities, etc., to avoid any contraindication of FES. All of the paraplegic patients had no pain sensation at their thighs, and were active and cooperative in this study. Each of them was given detailed explanation about every experimental procedure and any potential danger that could occur during experiments, such as skin irritation. All experiments were done with their written consents.

Torque Measurement and Muscle Stimulation

The subject was seated on a chair, developed by our research team, with the trunk at the upright position, and the knee angle was fixed at two different positions, 90° and 150°. A load cell (UB-K100, Jungsan, Inc., Seoul, Korea) was used for measuring the force F acting on the ankle when the knee extensor was stimulated. Equation (1) derived from the free body diagram (Fig. 1) was applied to calculate the knee extensor torque M :

$$M = FL_c + mgL_m \sin(q_k - 90^\circ) \quad (1)$$

The distance between the knee joint and the mass center of the leg, L_m , was estimated based on the anthropometric data [21], and utilized for determining the mass below the knee joint by the balance board technique [21]. L_c is the distance between the knee joint and the load cell, and q_k and g are the knee angle and the gravitational constant, respectively.

A reference electrode was attached to the patella (PA), and a round surface electrode (3M, Maplewood, MN, USA) with the radius of 1.5cm was positioned to the motor point of the muscle to be stimulated. The motor point was determined manually as the point that provided the maximum muscle contraction. We stimulated rectus femoris (RF) to measure the knee extensor torque, and gastrocnemius medialis (GM) was added in the standing/walking program.

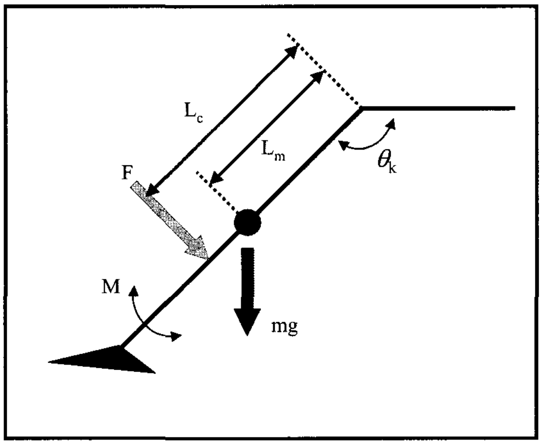


Fig. 1. Free-body diagram for computing the isometric knee extensor torque

Voluntary contraction of the healthy subjects' knee extensors was monitored for up to 2 minutes. On the other hand, electrical stimulation was delivered to the knee extensors of both the healthy subjects and the paralyzed patients. Monophasic rectangular pulse trains were used at the frequencies of 20Hz and 80Hz. The monophasic pulse train was adopted because it was reported to delay atrophy more than the biphasic pulse train [22], whereas it causes more inconvenience to healthy subjects and generates lower force, given the same stimulus parameters, than the biphasic pulse train [23]. The other two stimulus parameters representing the stimulation intensity, the pulse width and the pulse amplitude, were adjusted to provide approximately 25% of the maximum voluntary contraction (MVC) force of the healthy subjects. The reason was that the body weight can be balanced when the electrically stimulated knee extensors generate 25% of those of a healthy subject [24,25]. As for the healthy subject, the stimulation intensity was adjusted to generate the maximum force without pain. Two types of stimulation were employed: continuous and intermittent (10s on and 10s off) stimulations. The knee angle (q_k in Fig. 1) was varied to investigate how the fatigue characteristic depends on the muscle length. The subject was given a resting period for more than 10 minutes between two stimulation sessions.

Four different types of pulses were employed to investigate how the muscle torque and the muscle fatigue are affected by the waveform of the stimulation pulse: rectangular monophasic pulses (P_{mono}), 50%-balanced rectangular biphasic pulses ($P_{50\%}$), 25%-balanced rectangular biphasic pulses ($P_{25\%}$), and one-period sinusoidal pulses (P_{sine}), as shown in Fig. 2. Six healthy subjects and one paraplegic patient participated in this experiment. The average power of the constant-current (CC) pulse train was kept constant for each subject, and each pulse train was applied to the subject on eight different days to avoid any effect caused by previous simulations. The maximum peak torque and the fatigue index, defined as equation (2), were measured and averaged for each subject. The paired *t*-tests were done to each pair of the measurement groups, e.g., torques generated by P_{mono} and those by P_{sine}.

$$\text{Fatigue index (\%)} @ \frac{\text{Initial torque - Torque after 30s stimulation}}{\text{Initial torque}} \times 100 \quad (2)$$

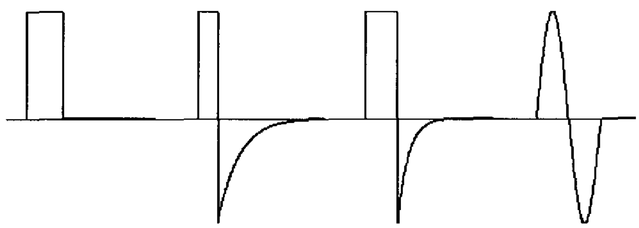
In all torque measurements, the torque was normalized with respect to the initial (maximum) value, and the results from different subjects were all averaged. Emphasis was given to the time instant when the knee torque decreased to 50% of the initial maximum value, which is called $T_{50\%}$ in this paper. In addition to $T_{50\%}$, we also employed the fatigue index. The subjects were told and trained not to move their upper extremities and not to voluntarily contract the leg muscles during the experiments.

One of the four paraplegic patients (the subject NS, male, age 37, 58kg, 173cm, L-1 incomplete, fall) started an FES standing/walking program. Throughout the program, reusable self-adhering electrodes (5cm×5cm, Neuron Technologies, Wolfeboro, NH, USA) were used for delivering CC monophasic rectangular pulse trains to vastus lateralis (VL), vastus medialis (VM), RF, GM, and gastrocnemius lateralis (GL). In this study we utilized a PC-based 8-channel CC transcutaneous stimulation system developed in our laboratory. Details about the system including the pros and cons are described in [26].

The FES standing was started when the knee extensor torque reached 20~25Nm that was enough to balance the subject NS's body weight [10]. Parallel bars were used for the first one month, and then replaced by a walker. The standing practice was done twice a week, 5~6 sessions a day. We tested three different combinations of the electrode position and the stimulus parameters for standing:

(a) RF(-) ⊕ PA(+), VL(-) ⊕ VM(+); 145~150V, 700~1,000s, 20~30Hz
(b) RF(-) ⊕ VL(+), VM(-) ⊕ PA(+); 145~150V, 700~1,000s, 20~30Hz
(c) RF(-) ⊕ VL(+); 147~150V, 700~1,000s, 20~30Hz
VM(-) ⊕ GM(+); 120~140V, 200~700s, 20Hz

Here, (+) and (-) denote the anode and the cathode, respectively



(a)P_{mono} (b)$P_{50\%}$ (c)$P_{25\%}$ (d)P_{sine}

Fig. 2. Stimulus waveforms selected in this study: (a) Monophasic rectangular waveform, (b) Unsymmetric balanced biphasic waveform, (c) Unsymmetric unbalanced biphasic waveform, and (d) Sinusoidal waveform

The standing posture was changed depending on whether GM was used or not. In case of (a) and (b), the waist was hyperextended such that the mass center of the whole body lies right above the ankle joint. When GM was used in (c), however, the mass center stayed right above or slightly in front of the knee joint since GM prevented dorsiflexion. In (c), the subject NS did not need to hyperextend his waist, and consequently the resulting standing posture was close to normal. When the reference electrode was placed right above PA, co-contraction of the hamstrings was observed, and, therefore, the reference electrode was moved in the proximal direction by 2~3cm.

Two types of posture switching [27,28] were implemented to extend the standing time. First, while standing, the knee extensors of the two legs were alternately stimulated and the subject NS moved his body weight to the stimulated leg. The other type was to stimulate RF and VL of one leg and VM and GM of the other leg at the same time, and to switch the stimulation, when the subject NS did not have to move his body weight.

Table 1. Combinations of the electrode position and the stimulus parameters for FES walking; (+) and (-) denote the anode and the cathode, respectively.

Combination	Electrode position	Pulse amplitude(V)	Pulse width(μs)	Frequency (Hz)
(a)	RF(−) ⊕ VL(+) & VM(−) ⊕ PA(+)	135~150	700~1,000	20
	& PN(−) ⊕ GL(+)	120~130	700	20
(b)	RF(−) ⊕ VL(+)	147~150	700~1,000	20
	& PN(−) ⊕ GL(+)	120~130	700	20
(c)	RF(−) ⊕ VL(+) & VM(−) ⊕ PA(+)	135~150	700~1,000	20
(d)	RF(−) ⊕ VL(+)	147~150	700~1,000	20

In addition to posture switching, the subject NS practiced to distribute his body weight evenly to two legs, to lift the legs utilizing his residual function, and to induce the flexion reflex by stimulating PN.

FES walking was started when the subject NS was able to stand up, sit down, and maintain the body balance freely enough with a walker. At first, the subject NS practiced reciprocal walking while a therapist manipulated the stimulator. Later, stimulation of the knee extensors was controlled by the subject NS himself. As mentioned above, the subject NS utilized the flexion reflex during the swing phase to avoid dragging the swinging leg by stimulating PN, and at the same time he kept practicing to lift the swinging leg for himself. The latter was possible since the exercise program enhanced the subject NS's leg lifting ability remarkably. Four different combinations of the electrode position were tested as shown in Table 1.

RESULTS AND DISCUSSION

When VL of the healthy subject was stimulated at 20Hz and 80Hz, the resulting $T_{50\%}$ was 174s and 58s, respectively, whereas MVC maintained more than 70% of the initial torque after 170s. The 80Hz stimulation showed much higher initial torque and much faster fatigue than the 20Hz stimulation. Stimulations of the paralyzed muscles showed $T_{50\%}$ equal to 12s and 46s at 80Hz and 20Hz. Intermittent stimulation maintained 70% (80Hz) and 88% (20Hz) of the initial knee torque at the time instant when the knee torque decreased to 50% in continuous stimulation at 80Hz. The above results confirmed that high-frequency and continuous stimulation causes faster fatigue than low-frequency and intermittent stimulation, respectively.

$T_{50\%}$ was 46s (20Hz) and 12s (80Hz) when VL was continuously stimulated at $q_k = 90^\circ$ whereas 42s (20Hz) and 8.5s (80Hz) at $q_k = 150^\circ$. That is, $q_k = 90^\circ$ resulted in higher fatigue resistance than $q_k = 150^\circ$. In MVC, however, we found the opposite, i.e., $T_{50\%}$ was 88s and 185s at $q_k = 90^\circ$ and $q_k = 150^\circ$, respectively, which meant that muscles at $q_k = 150^\circ$ were more fatigue resistant than at $q_k = 90^\circ$. The result from stimulated muscles is in good agreement of other experimental observations [29,30].

Fig. 3 shows the averaged peak torques normalized with respect to the torque generated by P_{mono}. The paired *t*-test results with the level of significance at 0.01 indicated that P_{sine} generated the largest torque and the other three pulse trains generated almost the same amount of torques in five out of the six healthy subjects, whereas, in one healthy subject and the paraplegic patient, the waveform of the pulse train made virtually no difference in the torque. This observation may be explained by employing the recruitment curve of each muscle as shown in Fig. 4. If a muscle is fully (i.e., supramaximally) stimulated, then all muscle fiber are recruited, and consequently the muscle force reaches the saturation level regardless of the waveform. Our observation suggested that the muscle force generated by partial stimulation may depend on the waveform, and that the slope of the increasing part of the recruitment curve is the largest in case of P_{sine}.

The fatigue index measurements (Fig. 5) showed that, in general, P_{sine} caused less fatigue than the other three pulse trains, and that the monophasic pulses caused less fatigue than the biphasic pulses. However, the fatigue measurement results were statistically less meaningful than the torque results, and more experiments are needed to confirm the results. Fig. 5 also shows and confirms that healthy voluntary contraction of a healthy muscle causes less fatigue than stimulation-induced contraction of a paralyzed muscle.

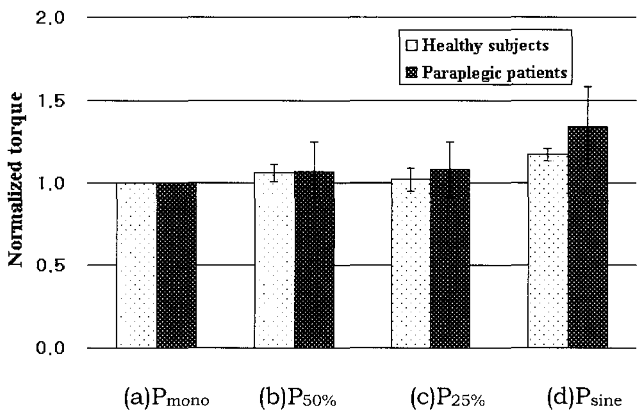


Fig. 3. Initial maximum knee torque normalized with respect to that generated by P_{mono} (a) Healthy subjects, and (b) Paraplegic patients

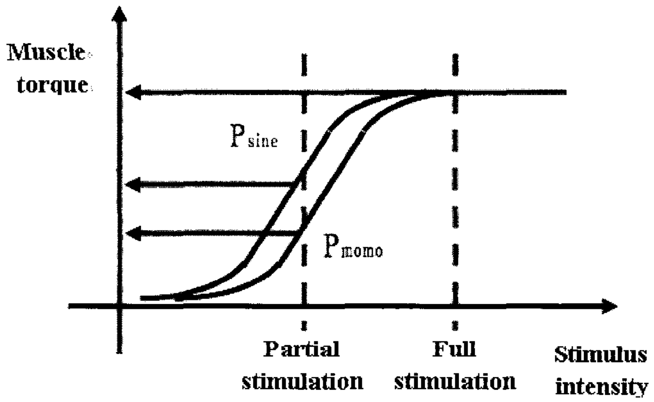


Fig. 4. Recruitment curves for two different waveforms

The subject NS's standing duration was about 40s at the beginning of the standing program. As far as the electrode position was concerned, (b) resulted in a longer standing duration and a larger knee torque than (a). When GM was stimulated as in (c), the standing duration was the longest and the standing posture was very close to that of a healthy person. When the knee extensors of the two legs were stimulated alternately for posture switching, the standing duration was 1 minute and 30s. On the other hand, the standing duration was prolonged up to 1 minute and 50s when two different muscle groups were stimulated alternately, i.e., at the second posture switching described in METHOD.

The subject NS's gastrocnemius showed less atrophy than his RF and vasti, and thus we could prolong the standing duration by stabilizing the ankle. Although gastrocnemius, which is a double-joint muscle, is known to be a knee flexor since it is located at the posterior side of the knee joint, it can accelerate the knee joint in the direction of extension depending on the position and the body-segmental parameters (see [12] for details). We believe that the subject NS's gastrocnemius accelerated his knee joint toward extension at the upright position, and, consequently, acted as a knee extensor. However, the stimulation intensity of GM needed adjusting at the beginning of every experiment because strong stimulation of GM often caused the heel-off.

The subject NS could walk for 5 minutes at the speed of 12m/min. Among the four combinations of the electrode position, (b) resulted in the best walking in terms of the joint angles. Stimulation of PN could lessen awkward walking by inducing the flexion reflex during the swing phase, whereas direct stimulation of the dorsiflexors may sometimes causes about 1-second latency. Also, the flexion reflex could prevent the legs from scissoring during the swing phase. Although such flexion reflex was of great help at the beginning of the walking program, NS could finally walk with his face facing toward the anterior direction without stimulation of PN since his physiological function to lift the leg was significantly improved through the exercise program. Stimulation of VL during the stance phase turned out to be better than stimulation of VM, because the knee flexion seemed to be prohibited by the prolonged contraction of VM at the swing phase.

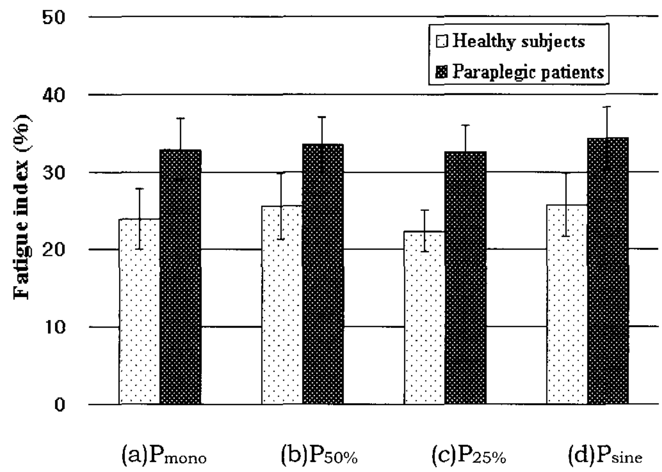


Fig. 5. Fatigue index (average ± standard deviation) for different waveforms (a) Healthy subjects, and (b) Paraplegic patients

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