

Interferential Electrical Stimulation Applied to the Soleus Muscle in Humans: Preliminary Study on the Relationship among Stimulation Parameters, Force Output, and Pain Sensation

Kohei Kozasa¹, Ryo Fujihara¹, Hiroaki Hirai¹, *Member, IEEE*, and Hermano Igo Krebs², *Fellow, IEEE*

Abstract—This study presents a method of providing deep muscle stimulation with an interferential current via surface electrodes, and analyzes the response of the skeletal muscles and the pain sensation experienced by subjects. Three healthy adult males were subjected to isometric contraction to generate the interferential current in the soleus muscle group, and the degree of muscle contraction by the plantarflexion torque and the pain sensation of the subjects were evaluated. Two square waves with different pulse widths were generated at the same time, and acted as the stimulation current with a relatively small amplitude of ± 16 mA. The experiments confirmed that a larger output torque could be obtained by generating interferential waves, compared to the condition where two identical waves (non-interferential current) were used. In addition, it was confirmed that the output torque and the pain sensation of the subjects decreased with the increase in the average frequency of the interferential current. When an interferential current was generated under the condition of interferential frequency $f_{Low} = 30$ Hz and average frequency $f_{High} = 50 \sim 100$ Hz with the proposed electrode arrangement, subjects produced the largest output without pain. The results can contribute to assisting ankle pushoff movement in walking during the rehabilitation of patients with central motor paralysis.

I. INTRODUCTION

The demand for neurorehabilitation technologies is growing, with an increase in the elderly population and age-related disorders. The majority of patients suffering central motor paralysis caused by stroke or spinal-cord injury are subjected to rehabilitation. Since the peripheral nerve cannot receive normal instructions from the central nerve, muscles cannot perform their motor functions despite possessing physiological excitability. Therefore, a method has been proposed in which an electric current directly stimulates the peripheral nerve or depolarizes the postsynaptic membrane to suppress muscle contraction, without going through the central nervous system. This method is known as functional electrical stimulation (FES) of the motor system [1].

¹Kohei Kozasa, Ryo Fujihara, and Hiroaki Hirai are with the Graduate School of Engineering Science, Osaka University, Toyonaka, Japan {kozasa, fujihara, hirai}@robotics.me.es, osaka-u.ac.jp

²Hermano Igo Krebs is with the Department of Mechanical Engineering, Massachusetts Institute of Technology, Cambridge, USA, the Department of Neurology, University of Maryland, School of Medicine, Baltimore, USA, the Department of Physical Medicine and Rehabilitation, Fujita Health University, Nagoya, Japan, the Institute of Neuroscience, Newcastle University, Newcastle upon Tyne, UK, the Wolfson School of Engineering, Loughborough University, Loughborough, UK, Programme Bioengineering & Biosciences, University Campus Bio-Medico of Rome, Rome, Italy, and the Department of Mechanical Science and Bioengineering, Osaka University, Osaka, Japan hikrebs@mit.edu

Currently, three types of electrode systems are used for FES: surface electrodes, needle electrodes, and implanted electrodes. The surface electrode system pastes a rubber electrode on the skin using a highly conductive adhesive. It is easy to attach to the skin. Although the selectivity of this system can be increased by adjustments through electrical contacts [2], it is only suitable for the surface motor nerves. The needle electrode system is composed of a guide needle and a wire electrode [1]. It is embedded at the motor point of the target muscle using a needle. The needle electrode system can stimulate deep muscles and small muscles, individually; however, the electrode always penetrates the skin. Thus, there is a risk of infection and of the electrode breaking. Furthermore, the procedure must be performed by a medical doctor. The implanted electrode is completely embedded inside the body, including the stimulator and receiver, similar to a cardiac pacemaker. It can reliably stimulate deep muscles and multiple muscles. However, many problems exist, such as the need for surgery for implantation and the risk of infection [1]. By contrast, the interferential current method is noninvasive and can stimulate deep muscles. This method involves synthesizing two different high-frequency waves by using multiple surface electrodes to generate a low-frequency interferential wave at the target point in-vivo. Since nerves become unresponsive when the stimulation frequency increases, if either one of the high-frequency waves is flowing, it will be difficult to cause muscle contraction. When the stimulation frequency increases, the skin impedance will decrease and we will be able to electrify the deep layers with less pain. The interferential frequency corresponds to the difference of the two input frequencies. Theoretically, adjusting the interferential frequency can stimulate only the muscles located at the point of interference [3]. If a large output can be obtained by using an interferential current that can stimulate deep muscles noninvasively, it will be useful in the rehabilitation field, for instance, in the use of the soleus muscle during the pushoff phase in walking. In research on therapeutic electrical stimulation (TES), there have been reports that dysmenorrhea and refractory chronic constipation of children have been improved by using the interferential current method [3] [4]. It has also been confirmed that a muscle contraction force as high as 50% of the maximum voluntary contraction can be obtained with an interferential current [5].

Many commercial products using surface FES have been put to practical use. The NESS L300 (Bioness Inc., USA)

is an assistive device acting on the ankle during walking. This product distinguishes gait phases using sensors and stimulates the tibialis anterior muscle during the swing phase using surface electrodes, to prevent patients with symptoms of drop foot from stumbling [6]. Most products acting on the ankle using surface FES assistive devices for walking are aimed at preventing drop foot by dorsiflexion assistance, and there are few products aimed at ankle pushoff assistance. The gastrocnemius muscle group, which is a group of surface muscles in the ankle joint plantarflexor, is easy to stimulate noninvasively; however, because it is a biarticular muscle affecting ankle- and knee-joint movement, it cannot produce the entire force for ankle pushoff. The soleus muscle is a mono-articular muscle, but it is difficult to stimulate it noninvasively because it is a deep muscle. These are the main problems of surface FES devices for ankle pushoff assistance.

This study explores the feasibility of the interferential current approach for the use of soleus muscle stimulation. When actually generating an interferential current in-vivo, attenuation due to bioelectrical impedance occurs and an ideal interferential current may not be generated [7]. The purpose of this study is to investigate the response of skeletal muscles and the subjective pain experienced by subjects whose soleus muscles were stimulated by interferential current. The specific methods employed are described next. First, we located the placement of the surface electrodes for generating the interferential current for the soleus muscle, using an ultrasound diagnostic system. Then, we prepared the two frequency parameters constituting the interferential current. The plantarflexion torque was measured as the output of the skeletal muscle when stimulating the ankle plantarflexor muscle group by the arranged electrode placement. A questionnaire was handed out after each trial, and the subjects were asked to score the sensation of pain according to given criteria.

II. MATERIALS AND METHODS

A. Participants

Three healthy adults males (168.0 ± 7.5 cm, 55.0 ± 8.7 kg, 22-23 years old, right-legged) volunteered for the experiment. None of them reported any history of neuromuscular disorders or injuries. The experiment was approved by the Ethical Committee of the Graduate School of Engineering Science, Osaka University, and all the subjects provided informed consent prior to inclusion.

B. Apparatus

The FES system consisted of a multichannel stimulus generator (STG 4008, Multi-Channel Systems, Inc., Germany) and a PC. The specifications for the device are as follows: maximum stimulation current ± 16 mA, maximum stimulation voltage ± 8 V, and time resolution $20 \mu\text{sec}$. The surface electrodes (PERFORMANCE, Compex Inc., USA) were attached on the skin to generate an interferential current into the soleus muscle group, and the response of the skeletal muscle to the stimulation was observed with a portable

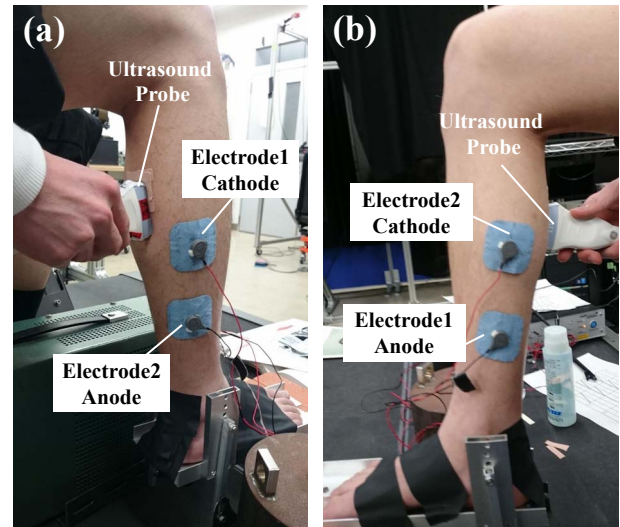


Fig. 1. FES electrodes displacement. (a) right-side electrodes. (b) left-side electrodes.

ultrasound system (LOGIQ P5, GE Healthcare Ltd., UK) and a force plate (ITR5018, Bertec Corp., USA).

C. Positioning of FES electrodes and ultrasound probe

While confirming the contraction of the soleus muscle upon electrical stimulation with an ultrasound diagnostic system, we searched for an electrode arrangement of four surface electrodes that could stimulate the soleus muscle of the right (dominant) leg. It is common to use a motor point pen when locating the electrode arrangement to stimulate the muscles of the surface layer. However, in our experiments, we mainly focused on the soleus, which is a deep muscle, making it difficult to search the motor point with a motor point pen. Thus, we opted to use an ultrasound probe, which we placed in such a position that images of the outer heads of the lateral gastrocnemius muscle and soleus muscle could be taken at the same time. The actual electrode arrangement and the position of the ultrasonic probe are shown in Fig. 1. The electrode positions normalized by the subjects' shank length were 0.472 ± 0.026 and 0.678 ± 0.005 for the cathode and anode of electrode 1, respectively, and 0.419 ± 0.019 and 0.675 ± 0.020 for the cathode and anode of electrode 2, respectively. The surface imaged by the ultrasound diagnostic apparatus is shown as a thick green line in Fig. 2.

D. Interferential current

Figure 3 shows a square-wave current with the amplitude A (16 mA) and pulse width T in the positive and negative phases. This waveform was adopted as the source for the interferential current stimulation. Compared to the stimulation waves commonly used, the square waves we adopted have very large pulse widths, because considerable stimulation intensity is needed to sufficiently stimulate the deep soleus muscle with large electrical impedance [8]. However, the maximum amplitude of the square wave used in this study was relatively small (16 mA).

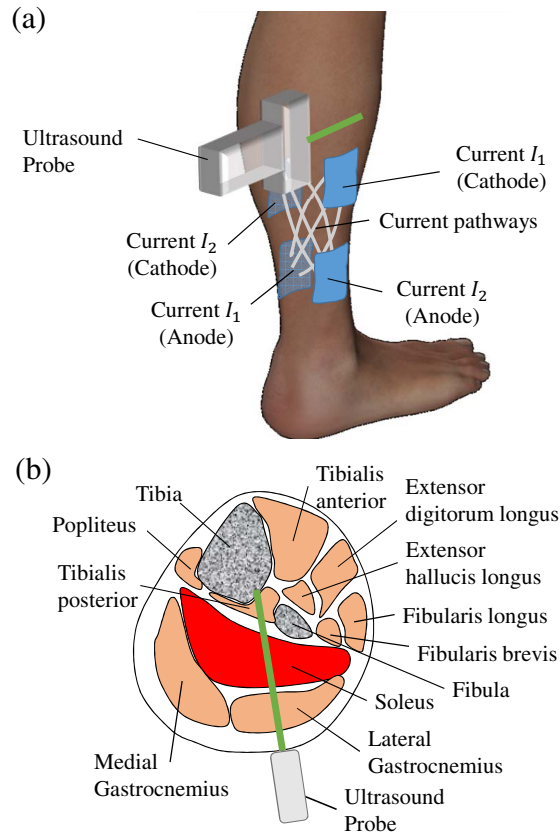


Fig. 2. Ultrasound experimental setup. (a) displacement of ultrasound probe and two pairs of surface FES electrodes. (b) target muscle for ultrasound imaging.

The Fourier series expansion of a square-wave current $I(t)$ shown in Fig. 3 is expressed as

$$I(t) = \frac{4A}{\pi} \sum_{n=1}^{\infty} \frac{1}{2n-1} \sin \frac{(2n-1)\pi}{T} t. \quad (1)$$

Equation (1) shows a series of trigonometric functions with frequencies that are odd multiples of the base frequency $f = \frac{1}{2T}$. Then, the combination of square-wave currents $I_1(t)$ and $I_2(t)$ with different pulse widths T_1 and T_2 can be expressed as

$$\begin{aligned} & I_1(t) + I_2(t) \\ &= \frac{4A}{\pi} \sum_{n=1}^{\infty} \frac{1}{2n-1} \left[\sin \frac{(2n-1)\pi}{T_1} t + \sin \frac{(2n-1)\pi}{T_2} t \right] \\ &= \frac{8A}{\pi} \sum_{n=1}^{\infty} \frac{1}{2n-1} \left[\sin \left\{ \frac{(2n-1)\pi}{2T_1} + \frac{(2n-1)\pi}{2T_2} \right\} t \right. \\ & \quad \cdot \cos \left\{ \frac{(2n-1)\pi}{2T_1} - \frac{(2n-1)\pi}{2T_2} \right\} t \left. \right]. \end{aligned} \quad (2)$$

Equation (2) is a series of interferential currents, that together form the product of trigonometric functions having high-frequency components and low-frequency components. In this paper, the high-frequency components of the base frequency of the interferential current (term corresponding to $n = 1$ in Eq. (2)) are denoted by $f_{High} (= \frac{1}{2}(\frac{1}{2T_1} + \frac{1}{2T_2}))$

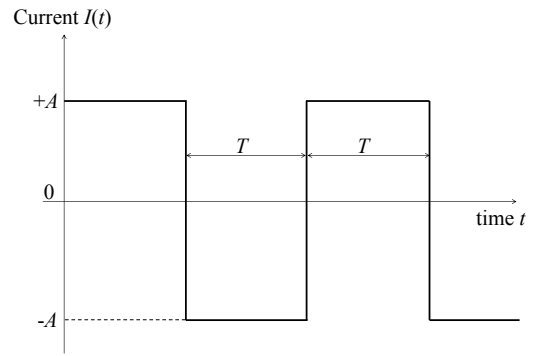


Fig. 3. Current stimulation waveform. A square wave has the same amplitude A and pulse width T in the positive and negative phases. This waveform is the source for the interferential current stimulation.

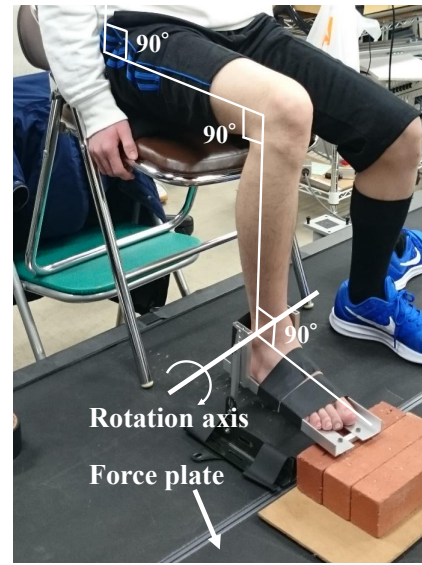


Fig. 4. Experimental setup. The subject sat in a chair to maintain his hip, knee, and ankle at a right angle. The instrument had a degree of freedom around the center of rotation of the ankle joint but this was prevented by a rigid block (isometric condition). The plantarflexion torque during electrical stimulation was measured by the force plate.

and the low-frequency components are denoted by $f_{Low} (= |\frac{1}{2T_1} - \frac{1}{2T_2}|)$. From Eq. (2), if two frequencies of the interferential current are determined, the pulse widths T_1 and T_2 of the square wave of the synthesis source can be obtained. The experimental conditions were determined using two frequencies (f_{High} and f_{Low}) of the interferential current generated using two pulse widths (T_1 and T_2) of source currents as parameters. Note that as a special case, the interferential frequency $f_{Low} = 0$ Hz (i.e., $T_1 = T_2$) yields non-interferential current with an (average) frequency $f_{High} = \frac{1}{2T_1}$.

E. Protocol

The subjects sat in chairs and their right (dominant) foot was fixed to the instrument with the hip, knee, and ankle all inclined at 90° . The instrument had freedom of rotation around the center of rotation of the ankle joint, but this was prevented by a rigid block. The subjects were asked to

TABLE I
EXPERIMENTAL CONDITIONS AND EXECUTION SEQUENCE
(EXPERIMENT 1)

Stim. order		f_{High}		
		10	20	30
Ascending	f_{Low}	0	1 ↓	3 ↓
		10	2 ↓	4 ↓
		20	-	5 ↓
		30	-	6 ↓
				7 ↓
Descending	f_{Low}	0	20 ↑	18 ↑
		10	19 ↑	17 ↑
		20	-	16 ↑
		30	-	15 ↑
				14 ↑
Random	f_{Low}	0	23	24
		10	27	21
		20	-	30
		30	-	26
				28

TABLE II
EXPERIMENTAL CONDITIONS AND EXECUTION SEQUENCE
(EXPERIMENT 2)

Stim. order		f_{High}					
		30	50	100	300	500	900
Ascending	f_{Low}	30	1	2	3	4	5
Descending	f_{Low}	30	12	11	10	9	8
Random	f_{Low}	30	16	17	15	13	18

perform maximum voluntary plantarflexion for 5 s under this isometric condition. The maximum value was adopted as a maximum voluntary contraction (MVC) torque to normalize the plantarflexion torque conducted by electrical stimulation. After the MVC torque measurement, an interferential current was generated in the triceps surae muscles of the subject's right leg, while the isometric plantarflexion torque generated by the right ankle joint was measured with a force sensor at a sampling rate of 1 kHz. Figure 4 shows the experimental setup. During the electrical stimulation, the lateral gastrocnemius and soleus muscle were also photographed by the ultrasound diagnostic system (Figs. 1 and 2).

Under this environment, two sets of experiments were performed.

Experiment 1: We investigated the difference between the skeletal muscle output of the square-wave (non-interferential) current and that of the interferential current. Table I shows the experimental conditions and execution sequence for Experiment 1. The execution sequence consisted of 30 conditions under different combinations of f_{Low} and f_{High} , and the sequence was divided into three groups according

(a) Non-Stimulated (b) Stimulated

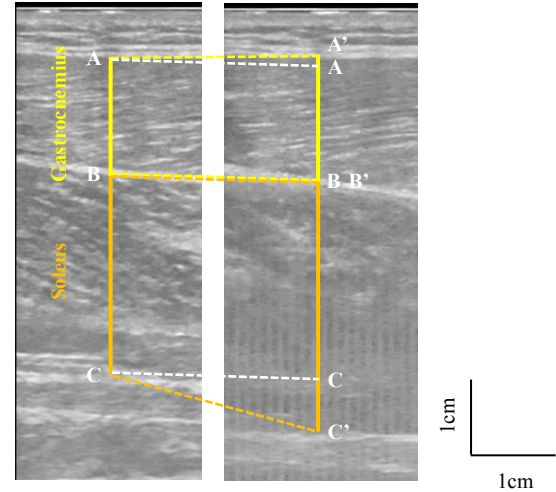
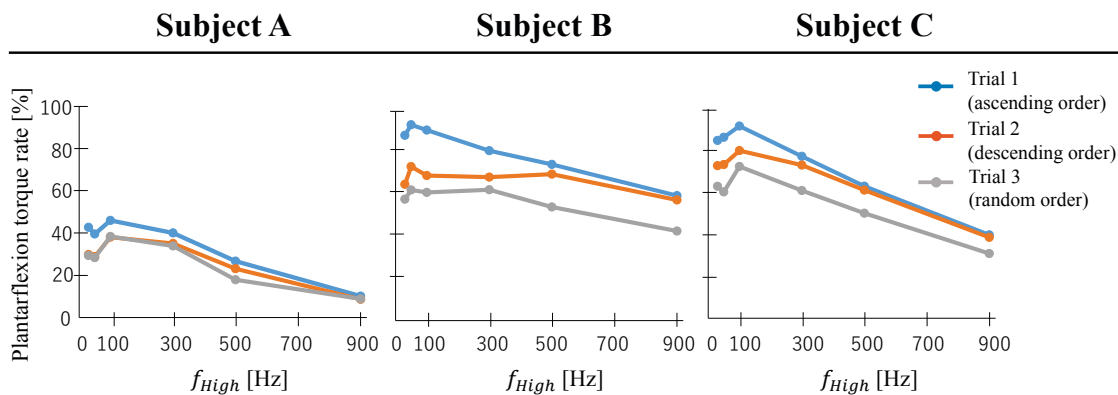
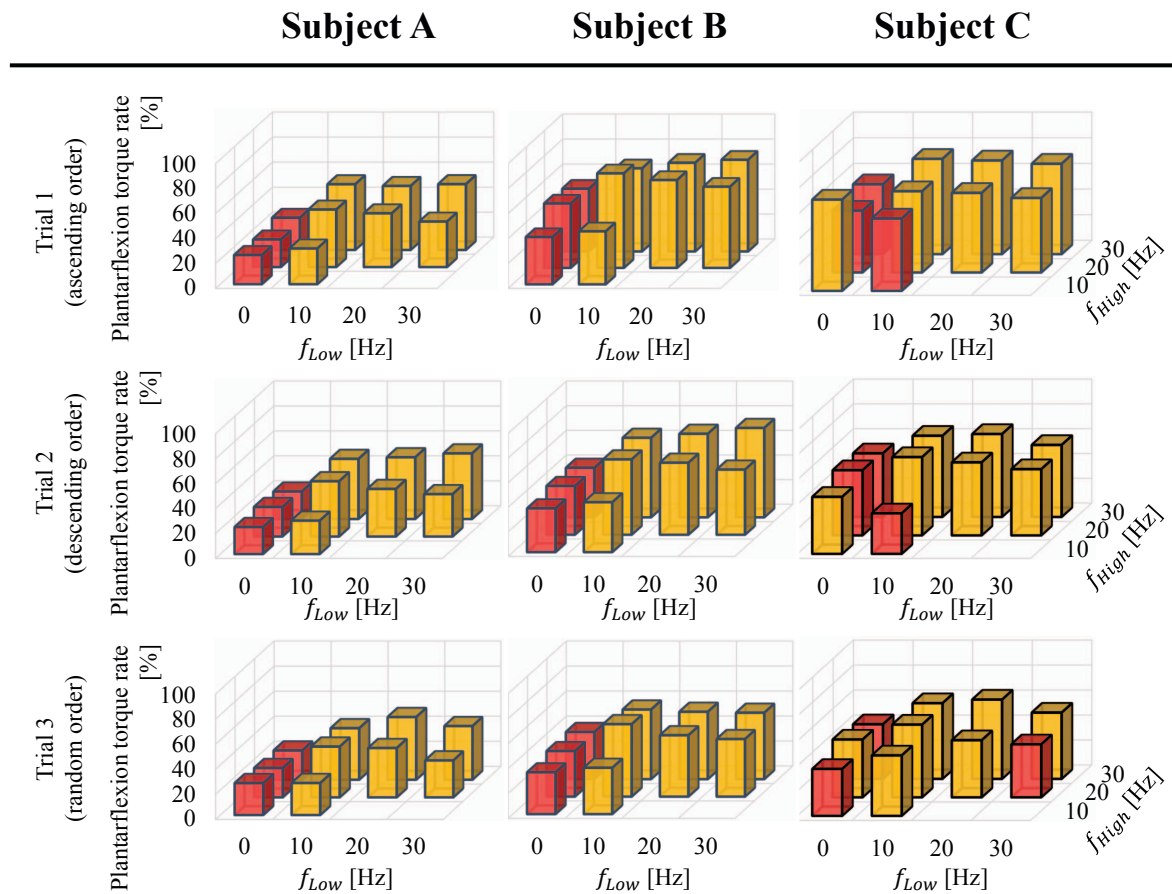


Fig. 5. Ultrasound image of lateral gastrocnemius and soleus muscle tissues. (a) No electrical current stimulation. (b) Interferential current stimulation with $f_{Low} = 30$ Hz and $f_{High} = 500$ Hz. The prime letters (A', B', and C') indicate the location of corresponding specific parts (A, B, and C) in the lateral gastrocnemius and soleus muscle tissues during interferential current stimulation. Each part was manually decided. Significant change in B-C (orange line) indicate that the deep muscle of soleus was selectively stimulated.

to stimulation orders. The ascending and descending stimulation orders were sequences that respectively involved an increasing and decreasing f_{Low} at a fixed f_{High} . The random stimulation order was an execution sequence chosen without any direction. Each number (1-30) in Table I indicates the execution order.

Experiment 2: We investigated the difference in the skeletal muscle output when fixing f_{Low} at 30 Hz and changing f_{High} (30, 50, 100, 300, 500, and 900 Hz). Table II shows the conditions for Experiment 2 and the three execution sequences, that is, ascending, descending, and random stimulation orders at the fixed f_{Low} .

Two sets of experiments were conducted three times under each condition. As shown as the execution sequences in Table I and II, we performed the experiments in the ascending order of f_{Low} - f_{High} pair during the first trial set, in descending order during the second trial set, and in a random order during the third trial set. In consideration of muscle fatigue caused by electrical stimulation, the stimulus duration was set to 500 ms on each trial. An interval of three minutes was provided after each trial. Experiment 2 was conducted one week after Experiment 1. Furthermore, for all conditions of Experiment 1 and Experiment 2, a questionnaire was provided to the subjects at the end of each trial, and the pain experienced as a result of electrical stimulation was evaluated in terms of five levels. The following indicators were used to score the pain level: 0, no sensation; 1, felt a weak stimulus but no muscle contraction sensation; 2, felt muscle contraction without pain; 3, felt muscle contraction with acceptable pain; 4, felt muscle contraction with unacceptable pain.



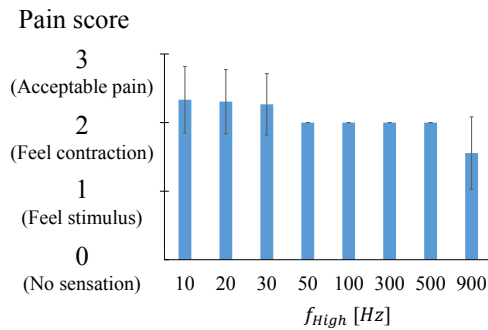


Fig. 8. Mean and standard deviation of pain score for all trials of the three subjects. The horizontal axis shows f_{High} , and the vertical axis shows the pain score. Pain scores depend on the average frequency f_{High} . Subjects tended to feel more comfortable as the average stimulation frequency increased.

III. RESULTS AND DISCUSSION

Hodges et al. examined the relationship between electromyography and muscle thickness and confirmed a positive correlation [9]. As shown in Fig. 5, the muscle thickness of the soleus muscle selectively increased during interferential current stimulation, compared to the thickness without stimulation. This result suggests that the soleus muscle is only activated by flowing two sets of current with different frequencies in the proposed electrode arrangement. Note that the lateral gastrocnemius muscle will activate if the interferential current stimulation is not used.

Furthermore, under the condition of $f_{Low} = 0$, that is, when the two square waves have the same pulse widths, we can see that the output torque has a tendency to decrease when compared to the other conditions (Fig. 6). Thus, interferential current can produce a larger output than the output from using ordinary (non-interferential) current. This result indicates the usefulness of the interferential current method, in terms of the output torque.

It was also confirmed that the pain score decreases as f_{High} increases (Fig. 8). The frequency characteristic of the biological tissue impedance is considered to be one of the factors contributing to this matter. Zheng et al. collected skeletal muscles of five types of mammals and chickens and measured the longitudinal and lateral electrical impedances in the range of 1 Hz to 1 MHz, to show that the impedance decreased as the frequency increased [10]. Rosell et al. measured the electrical impedance of human skin, in the range of 1 Hz to 1 MHz, and reported the same results as Zheng et al. [11].

The output torque peaks appeared around $f_{High} = 50 \sim 100$ Hz, after which the plantarflexion torque decreased with the increase in f_{High} , for all three subjects (Fig. 7). Increasing the frequency means that the stimulation time of one pulse is shortened, and one factor of the decrease in plantarflexion torque with the increase in f_{High} is the lack of stimulation intensity required to drive the muscle fiber.

To contract the muscle under the damaged central nervous system, it is necessary to depolarize the transverse tubule of the muscle fibers by applying some external stimulus.

A typical method is to induce a depolarization potential by exciting the axis of the motor neuron by electrical stimulation and changing the ion permeability of the postsynaptic membrane with a biochemical mechanism [12]. Another method provides the potential difference necessary for the depolarization of a transverse tubule directly from the outside by electric stimulation; however, in this case, much more energy than that required for exciting peripheral nerves is necessary [1], [13]. We used both nerve excitation and direct transverse tubule depolarization, up to approximately $f_{High} = 100$ Hz, but it seems that the decrease in stimulation intensity arising from the increase in stimulation frequency caused the contraction by nerve excitation only. The decrease in stimulation intensity also contributed to a reduction in subjective pain reported.

When an interferential current was generated from square waves under the condition of $f_{Low} = 30$ Hz and $f_{High} = 50 \sim 100$ Hz, with a relatively small amplitude of stimulation current (16 mA) using the proposed electrode arrangement, the subjects gained the largest output without pain. These results can contribute to the development of surface FES technology to assist the rehabilitation of ankle pushoff movement in paralytic patients.

REFERENCES

- [1] M. R. Popovic, K. Masani, and S. Micera, "Functional electrical stimulation therapy: recovery of function following spinal cord injury and stroke," *Neurorehabilitation Technology* (D. J. Reinkensmeyer and V. Dietz Eds.), Springer, 2nd ed., pp. 513-532, 2016.
- [2] A. Kuhn, T. Keller, S. Micera, and M. Morari, "Array electrode design for transcutaneous electrical stimulation: a simulation study," *Med. Eng. Phys.*, vol. 31, no. 8, pp. 945-951, 2009.
- [3] G. C. Goats, "Interferential current therapy," *Br. J. Sports Med.*, vol. 24, no. 2, pp. 87-92, 1990.
- [4] N. Tugay, T. Akbayrak, F. Demirtürk, I. C. Karakaya, O. Kocaacar, U. Tugay, M. G. Karakaya, and F. Demirtürk, "Effectiveness of transcutaneous electrical nerve stimulation and interferential current in primary dysmenorrhea," *Pain Med.*, vol. 8, no. 4, pp. 295-300, 2007.
- [5] J. Chase, V. J. Robertson, B. Southwell, J. Hutson, and S. Gibb, "Pilot study using transcutaneous electrical stimulation (interferential current) to treat chronic treatment-resistant constipation and soiling in children," *J. Gastroenterol. and Hepatol.*, vol. 20, no. 7, pp. 1054-1061, 2005.
- [6] J. M. Hausdorff and H. Ring, "Effects of a new radio frequency-controlled neuroprosthesis on gait symmetry and rhythmicity in patients with chronic hemiparesis," *Am. J. Phys. Med. Rehabil.*, vol. 87, no. 1, pp. 4-13, 2008.
- [7] V. Robertson, A. Ward, J. Low, and A. Reed, *Electrotherapy explained*, 4th edition, Elsevier, 2006.
- [8] O. Lagerquist and D. F. Collins, "Influence of stimulus pulse width on M-waves, H-reflexes, and torque during tetanic low-intensity neuromuscular stimulation," *Muscle Nerve*, vol. 42, no. 6, pp. 886-893, 2010.
- [9] P. W. Hodges, L. H. Pengel, R. D. Herbert, and S. C. Gandevia, "Measurement of muscle contraction with ultrasound imaging," *Muscle Nerve*, vol. 27, no. 6, pp. 682-692, 2003.
- [10] E. Zheng, S. Shao, and J. G. Webster, "Impedance of skeletal muscle from 1 Hz to 1 MHz," *IEEE Trans. Biomed. Eng.*, vol. 31, no. 6, pp. 477-481, 1984.
- [11] J. Rosell, J. Colominas, P. Riu, R. Pallas-Areny, and J. G. Webster, "Skin impedance from 1 Hz to 1 MHz," *IEEE Trans. Biomed. Eng.*, vol. 35, no. 8, pp. 649-651, 1988.
- [12] M. L. Latash, *Neurophysiological basis of movement*, second ed., Human Kinetics, 2008.
- [13] P. H. Peckham and J. S. Knutson, "Functional electrical stimulation for neuromuscular applications," *Annu. Rev. Biomed. Eng.*, vol. 7, pp. 327-360, 2005.