Development of a prototype of portable FES rehabilitation system for relearning of gait for hemiplegic subjects

Takashi Watanabe , Shun Endo, Ryusei Morita

Graduate School of Biomedical Engineering, Tohoku University, Sendai 980-8579, Japan

✉ E-mail: nabet@bme.tohoku.ac.jp

Published in Healthcare Technology Letters; Received on 19th May 2016; Revised on 28th July 2016; Accepted on 3rd August 2016

This study aimed at developing a prototype of portable FES rehabilitation system for relearning gait pattern of healthy subjects, which can measure gait information during walking applying electrical stimulation for foot drop correction or providing timing information. A gait event detection method using an inertial sensor attached on the foot was determined based on gait of healthy subjects from simultaneous measurements with pressure sensors. From the result of comparing the detected gait event timings with EMG signal of the tibialis anterior muscle during walking of healthy subjects, the toe off and the foot flat timings detected by the inertial sensor were suggested to be useful to determine the stimulation timing for the foot drop correction. The gait event detection method was implemented in a prototype of portable FES rehabilitation system consisting of an 8-inch tablet-type device, 2 inertial sensors and an electrical stimulator. The portable system was examined with hemiplegic subjects under the conditions of FES foot drop correction and inducing voluntary effort to develop ankle dorsiflexion at the timing given by electrical stimulation with small stimulation intensity. The system was considered to be useful for gait rehabilitation of hemiplegia using FES foot drop correction or inducing voluntary effort.

1. Introduction: Foot drop correction improves significantly gait of hemiplegic subjects [1, 2]. For this purpose, while ankle foot orthosis (AFO) is widely used, functional electrical stimulation (FES) has also been shown to be effective for foot drop correction of hemiplegic subjects during gait [1–3]. Since FES has therapeutic effects in addition to restoring function of ankle dorsiflexion, FES can be an effective tool for gait rehabilitation for hemiplegic subjects [4, 5].

Voluntary effort to develop dorsiflexion at the timing given by low intensity electrical stimulation during walking were suggested to improve gait of a hemiplegic subject, showing similar results to those of FES foot drop correction [6]. That is, both of FES foot drop correction and the voluntary effort to develop dorsiflexion changed not only ankle dorsiflexion, but also other joint movements of the paralysed and non-paralysed sides during walking. Angle range of thigh inclination angle and stride time of the paralysed side were also changed. The results were obtained by measurements with inertial sensors during walking applying electrical stimulation. These suggest that gait information during rehabilitation training has to be measured in order to evaluate rehabilitation effects.

Generally, electrical stimulation for the foot drop correction is applied during the swing phase detected by foot switches [7]. On the other hand, inertial sensors have been studied on gait event detection [8], and some studies aimed for using inertial sensor in triggering electrical stimulation for foot drop correction [9, 10]. However, there are few studies that tested clinically the gait events detected by inertial sensors in applying electrical stimulation for foot drop correction [7, 11]. In addition, duration of the foot flat was not detected by an inertial sensor in most of studies. Although Sabatini *et al.* detected it by an inertial sensor, examination was not performed in over-ground walking, but in treadmill walking [12].

One of goals of gait rehabilitation can be regarded as relearning gait pattern of healthy subjects. For this purpose, it is desirable to apply electrical stimulation at timings close to muscle activities during gait of healthy subjects. However, relationship between the activity of the muscle that produce ankle dorsiflexion and the swing phase detected by an inertial sensor has not been validated. Although many studies used heel off timing as the beginning of electrical stimulation for foot drop correction [1, 4, 11, 13], it is considered that the timing is earlier than muscle activation for dorsiflexion.

This Letter aimed at developing a portable FES rehabilitation system that can be used in training for relearning gait pattern of healthy subjects, which can measure gait information during walking applying electrical stimulation for foot drop correction or timing information. First, a detection method of gait events using an inertial sensor attached on the foot was determined based on gait of healthy subjects. Then, the relationship between activity of the tibialis anterior muscle and gait event timings detected by the inertial sensor was compared. The timing of applying electrical stimulation is determined from these results, by which electrical stimulation is applied when foot movement pattern during gait is similar to those of healthy subjects. The developed prototype system was examined in applying electrical stimulation at the detected timings with hemiplegic subjects and in using for gait training under the conditions of FES foot drop correction and applying timing information to induce voluntary effort.

2. Gait event detection method based on gait of healthy subjects: A gait event detection method was developed based on gait pattern of healthy subjects, because the objective of gait rehabilitation using the system was to relearn gait of healthy subjects. Therefore, a gait event detection method with a wireless inertial sensor attached on the foot was developed based on gait of healthy subjects.

2.1. Experimental method: Walking movements were measured with nine healthy subjects (male, 21–24 years old). The subjects walked six times for 30 m straight way at three different walking speeds, moderate, slow and fast, determined by subjects themselves. The movements were measured simultaneously with the lower limbs motion measurement system using inertial sensors (WAA-010, Wireless Technologies, Inc.) [14] and pressure sensors (PH-462, PH-463, PH-464, DKH Co., Ltd.). The inertial sensors were attached by using stretchable band, which measured signals of a 3-axis accelerometer and 3-axis gyroscope with a sampling frequency of 100 Hz. The inertial sensor signals were transmitted to PC via Bluetooth. The signals measured with the sensors attached on the foot and the shank of the left side were analysed. The pressure sensors were attached on the left foot and their signals were measured with a sampling frequency of 100 Hz.

Pressure sensors were attached to four sections of outsole of a shoe as shown in Fig. 1 in order to obtain reference signals of gait event timings. Section A was the heel section with a square pressure sensor. Section B is rear sole section with two small circle pressure sensors. Section C is front sole section with four small circle pressure sensors. Section D is toe section with square pressure sensor. Each gait event timing was detected by the pressure sensors as follows:

Foot flat (FF): sections B and C are ON.

Heel off (HO): section B is OFF and section C is ON after the FF.

Toe off (TO): all section is OFF after section D is ON.

Initial contact (IC): section A is ON after the TO.

2.2. Results: The number of analysed strides were 1051, 1175 and 844 for the moderate, slow and fast speed walking, respectively. First, measured strides were divided by the IC into each stride. The IC was detected by longitudinal acceleration measured with the inertial sensor attached on the shank with a threshold value of -1200 mg [11]. Then, time of each stride was normalised to its gait cycle (GC) time between times of the IC, and measured signals were averaged. Unstable strides were removed manually by checking all the angle waveforms. Fig. 2 shows the average waveform of angular velocity of the foot with average gait event timings detected by pressure sensors. Table 1 shows average values of angular velocities of the foot at the gait event timings detected by pressure sensors. From these results, a detection method of gait event timings using inertial sensor attached on the foot was determined as follows:

HO: the time that angular velocity exceeds a threshold value.

TO: the time of the maximum value of angular velocity after the HO.

IC: the time of zero cross from the negative to the positive values of angular velocity after the TO.

FF: the time that the angular velocity is below a threshold value after the IC.

The threshold values to detect the HO and the FF were set as 19.0° and 91.3° from Table 1, respectively, considering that walking speed of motor disabled subjects were smaller than those of moderate speed walking of healthy subjects, since the angular velocities were different depending on walking speed.

Time difference of detected gait events between pressure sensors and the inertial sensor are shown in Table 2. In Table 2, the IC timing detected by the longitudinal acceleration measured with the inertial sensor attached on the shank with a threshold value of -1200 mg and the FF timing with the threshold of 25.0° , which were determined by trial and error process [11], are also shown.

Fig. 1 Locations of pressure sensors attached on outsole of a shoe

Positive values mean that gait event timings detected by the inertial sensor are earlier than reference values. The HO was detected at almost same timing. Timings of the TO and the IC could be detected by foot angular velocity about two sampling intervals earlier than pressure sensors. Detection of the FF timing was about four sampling intervals earlier with FF(91.3) and six sampling intervals later with FF(25.0) than pressure sensors.

2.3. Detection test with gait data of hemiplegic subjects: Since gait patterns of hemiplegic subjects are different from healthy subjects, the gait event detection method was tested preliminary using data measured with two hemiplegic subjects (right hemiplegia, males, 50 years old). In the measurement, each subject walked at moderate walking speed determined by himself with a cane in his left hand without support for the ankle joint such as AFO on 10 m straight way on level floor.

Here, two detection methods of the FF shown in Table 2 were applied. That is, method 1 used FF(91.3) and method 2 used FF(25.0). Detection methods of the HO, the TO and the IC were the same for both methods, which used angular velocity of the foot. Considering practical applications, time-out was set between detecting gait events in order to make appropriate detection by stopping for a short time after misdetection during walking. That is, gait event detection is initialised to wait the HO detection after detecting the time-out.

Detection rate is shown in Table 3. Although method 1 did not make misdetection for the HO, the TO and the IC timings, the FF timings were not detected due to small angular velocity after the IC. Method 2 showed higher detection rate than method 1. That is, decreasing the threshold value improved FF detection. Some misdetections were caused by the time-out set in the detection algorithm because of unstable foot movements during gait of the hemiplegic subject.

Foot angular velocity was useful because detection rates were more than 90%. Some misdetection can be provided to subjects as information of difference of gait from healthy subjects. Since there is a possibility of improving unstable gait by using electrical stimulation or voluntary effort, it is expected that the detection rate increases. Although the FF timing detected by FF(91.3) caused high rates of misdetection, decreasing the threshold value improved the FF detection. Therefore, the threshold value for the FF detection can be determined considering level of paralysis of subjects.

3. Validation test of stimulation timing: The gait event timings detected by the inertial sensor were compared with the timing of muscle activity for ankle dorsiflexion in order to determine stimulation timing for foot drop correction.

3.1. Method: Electromyogram (EMG) signals of the tibialis anterior that develops ankle dorsiflexion were measured simultaneously with an inertial sensor attached on the foot during walking with

Fig. 2 Average waveform of foot angular velocity (sagittal plane) and average gait event timings detected by pressure sensors (slow speed walking of healthy subjects)

Table 1 Average values of angular velocities of the foot at gait event timings detected by pressure sensors for different walking speeds (deg/s)

	Slow	Moderate	Fast
HO	19.0 ± 5.8	26.1 ± 8.8	21.5 ± 6.2
TO	264.2 ± 85.8	344.4 ± 108.3	482.3 ± 107.2
IC	-3.6 ± 59.5	-7.5 ± 95.3	-80.5 ± 122.3
FF	91.3 ± 40.1	128.2 ± 51.5	179.7 ± 83.7

ten healthy subjects (male, 21–24 years old). The subjects walked six times on 30 m straight way with a moderate walking speed determined by themselves.

EMG signals were measured with a sampling frequency of 1 kHz. Each EMG signals were full-wave rectified, normalised by its amplitude of EMG signals during the maximum voluntary contraction, filtered by Butterworth low-pass filter with the cut off frequency of 10 Hz, and resampled at 100 Hz in accordance with sampling frequency of inertial sensor signals. The processed EMG signals were divided into the signal for each stride and time of each stride were normalised by duration between the IC timings detected by angular velocity of the foot measured with the inertial sensor.

3.2. Results: Totally 1184 strides were analysed. Averaged processed EMG signal is shown in Fig. 3. Gait event timings shown in Fig. 3 were average values, in which 0% of the GC shows the IC and the FF detected by FF(25.0) is shown. Muscle activity was detected by amplitude larger than the minimum value of the EMG amplitude plus two times of standard deviation, which is shown by the shaded area with the threshold shown by broken line in Fig. 3. The tibialis anterior was activated between 0 and 14%GC, 63 and 80%GC and 92 and 100%GC. The FF, the HO and the TO timings detected by the inertial sensor were 11.2 ± 2.0%GC at the FF, 43.6 ± 2.5%GC at the HO and 63.0 ± 2.0%GC at the TO.

The beginning of EMG activity was almost same as the TO timing detected by the sensor. The difference of the FF timing detected by the inertial sensor from the end of EMG activity was about 3%GC. From these results, the TO and the FF timings is considered to be useful for detecting electrical stimulation timing in training of relearning gait of healthy subjects. However, the FF timing detected by FF(25.0) may be early for stopping electrical stimulation.

4. Test of a prototype FES rehabilitation system: The gait event detection method was implemented in a prototype of FES rehabilitation system and the system was examined in using for gait training under the conditions of FES foot drop correction and applying timing information to induce voluntary effort.

4.1. Outline of the prototype system: A prototype of portable FES rehabilitation system consisting of an 8 inch tablet-type device (Diginnos DG-D08IW, Windows 10 Home), two inertial sensors (WAA-010, CRESCO Wireless, Inc.) and a custom-made electrical stimulator was developed (Fig. 4). The inertial sensors and the stimulator are connected to the tablet-type device via

Table 2 Time differences of detected gait event timings between pressure sensors and an inertial sensor with healthy subjects (slow speed walking) (ms)

HO	TO	IC (foot)	IC (shank)	FF (91.3)	FF (25.0)
-2 ± 54	+21 ± 64	+16 ± 33	-55 ± 53	+42 ± 72	-56 ± 58

Table 3 Results of stride detection in 10 m walking of two hemiplegic subjects

Trial	Method 1		Method 2	
	Sub A	Sub B	Sub A	Sub B
1	12/12	6/19	12/12	16/19
2	11/11	5/19	11/11	17/19
3	7/12	16/18	11/12	17/18
4	11/12	11/18	12/12	18/18
5	6/13	15/16	10/13	16/16
6	8/12	15/16	11/12	12/16
Total	55/72	68/106	67/72	96/106
Rate, %	76.4	64.2	93.1	90.6

Bluetooth. Inertial sensors measure 3-axis components of acceleration and angular velocity with a sampling frequency of 100 Hz. Basically, the FES rehabilitation system can be applied for gait training with FES foot drop correction by using an inertial sensor attached on the paralysed foot. However, this system has an option of using the other inertial sensor for measurement of movements of another segment of the paralysed side or non-paralysed side.

Electrical stimulation for foot drop correction was applied to the tibialis anterior and the common peroneal nerve simultaneously by single channel stimulation using biphasic pulses. Stimulation frequency and pulse width was set to 30 Hz and 0.3 ms, respectively, which can be adjusted. Stimulation timing was determined by the TO and the FF timings detected by the inertial sensor attached on the foot, in which FF(25.0) was used for the FF detection because FF timing detected by FF(25.0) was closer to the end of muscle activity than FF(91.3).

4.2. Experimental method: The prototype system was applied to two subjects with right hemiplegia (males, 50 and 80 years old). Inertial sensors were attached on both feet by the same way as the measurement with healthy subjects. The subjects were instructed to walk freely at their moderate speed in a building or outside of the building. They used their canes during walking and had a rest several times in the measurements.

Walkings were measured by using two stimulation intensities: FES and vL-FES (voluntary effort with low intensity FES) conditions. Under the FES condition, subjects walked with FES foot drop correction using enough large stimulation intensity to develop ankle dorsiflexion. Under the vL-FES condition, they walked with voluntary effort to develop ankle dorsiflexion at the timing provided by electrical stimulation with small stimulation

Fig. 3 Normalised processed EMG signal of the tibialis anterior during gait. Gait event timings detected by the inertial sensor are also shown

Fig. 4 Prototype of portable FES rehabilitation system

intensity to perceive it, in which the small stimulation intensity could hardly develop ankle dorsiflexion.

Steady-state strides were analysed by removing the first two and the last two strides from each measurement trial. Strides before and after the non-detected stride (mis-detection) were also removed. Foot inclination angles were estimated by using Kalman filter from the data measured with the inertial sensors [15]. Inclination angles at the sitting position under the steady-state foot flat condition were measured before the walking test and the angles were set as 0°. Positive values of the angle means that the toe rises upward.

4.3. Results: The number of detected and misdetected strides during walking measurements with the portable system and the number of analysed strides are shown in Table 4. In the analysis, measured strides were divided by the detected IC into each stride. Then, time of each stride was normalised to its GC time between times of the IC, and averaged. Unstable strides were removed manually by checking all the angle waveforms. Detection rates under the vL-FES and the FES conditions were 93.6 and 95.3% for subject B, and 96.0 and 97.8% for subject C, respectively.

Analysed results of both subjects are shown in Fig. 5. Measured inclination angles, gait event timings and stride time were compared between vL-FES and FES conditions for the first 50 and the last 50 strides. For the first 50 strides, both subjects showed larger inclination angles under the FES condition than the vL-FES condition. The foot of the paralysed side contacted to the ground at the heel under the FES condition. However, inclination angles were similar between the FES and the vL-FES conditions for the last 50 strides. The foot inclination angle at the IC decreased under the FES condition. Since the both subjects have not used FES frequently, it is considered that muscle fatigue occurred in about 20 min FES walking.

As shown in Section 3.2, gait event timings of healthy subjects were about 11%GC at the FF, 44%GC at the HO and 63%GC at the TO. In the measurements with healthy subjects, stride time was about 1.2 s. The HO timings of both hemiplegic subjects were later than those of healthy subjects. It is found that gait

event timings of the vL-FES condition were delayed in comparison to the FES condition slightly in the first 50 cycles. Variations of the timings were large with subject B. Although the FF of subject B was almost same as healthy subjects, those of subject C were earlier for the last 50 strides. Stride time varied largely with the vL-FES condition for the first 50 strides and the time of both conditions decreased for the last 50 strides with both subjects.

5. Discussions: A method of detecting stimulation timing for foot drop correction with the inertial sensor was determined based on gait of healthy subjects and compared with muscle activity during walking, although there are differences in gait movement and gait event timings between healthy and hemiplegic subjects. This was because the objective of rehabilitation training using the system was to relearn gait of healthy subjects. The TO timing detected by inertial sensor was almost same as the beginning of activity of the tibialis anterior, while the HO was detected about 20%GC earlier than the muscle activity. Although some studies used the HO as the beginning of electrical stimulation for the foot drop correction [1, 4, 11, 13], it is considered to be early for realising healthy gait or providing its information. On the other hand, the FF timing detected by FF(25.0) with an inertial sensor was a little earlier than the end of the muscle activity. As shown in Table 2, the FF was detected by FF(25.0) about 60 ms later than pressure sensors. Since FF(91.3) detected the FF earlier than FF(25.0), FF(91.3) stops electrical stimulation earlier than FF(25.0). Therefore, FF(25.0) are considered to be preferable. Time difference between the end of EMG activity and the FF timing detected by FF(25.0) was about 3%GC, which is about 30–60 ms for stride time of 1–2 s. That is, electrical stimulation can be applied for the duration of activity of the tibialis anterior muscle by adding 1 or 2 pulses after the FF(25.0) detection.

As a gait event detection method, method 2 using FF(25.0) was implemented in the prototype system. The first reason of this was because the FF timing detected by FF(25.0) was closer to the end of muscle activity than FF(91.3). However, the FF timings detected by the inertial sensor were earlier than the end of the activity of tibialis anterior muscle, although FF(25.0) detected the FF timing later than pressure sensors. This shows that the tibialis anterior muscle activated after ankle dorsiflexion during swing phase and initial contact and in the FF duration. These activations of the tibialis anterior are considered to be for the loading response realising the heel rocker function [16]. For rehabilitation of relearning of gait, providing information of different roles of muscle activation such as dorsiflexion during swing and the loading response after the heel contact would be useful.

The other reason of using method 2 was that stride detection rate was higher than method 1. As shown in Table 3, stride detection rates of method 1 were lower than 80% because of mis-detection of the FF caused by small angular velocity near the FF with hemiplegic subjects. In the rehabilitation using the developed system, relearning of gait is considered to be difficult at the early stages of rehabilitation, if mis-detection rate is high. Therefore, small threshold value that could detect the FF with more than 90% detection rate was implemented. After improvement of gait movements, the FF(91.3) that uses larger threshold value as similar to healthy subjects can be used. Together with providing various gait information, this have to be studied for the next stage of rehabilitation.

The developed portable FES rehabilitation system is considered to be effective for gait training with a therapist and in chronic phase of hemiplegia. It is considered that trunk posture and thigh movements can be instructed by tapping its segment by a therapist during gait training. However, it is hard to give instruction of movements of the foot during walking. The developed portable system can be used for instructing the foot or ankle movements in gait training with a therapist. The developed device could make foot contact to the ground at the heel of the paralysed side under the

Table 4 Number of strides detected by the developed portable system and of analysed strides

		Sub B	Sub C
vL-FES	detect	781	288
	mis-detect	53	12
	analysed	663	237
FES	detect	769	622
	mis-detect	38	14
	analysed	686	586

Fig. 5 Analysed data measured during walking of hemiplegic subjects with the developed portable system. Left: foot inclination angle, centre: gait event timing, right: stride time

a First 50 strides of subject B
b Last 50 strides of subject B
c First 50 strides of subject C
d Last 50 strides of subject C

FES condition. Variations of stride time and gait event timings increased under the vL-FES condition in the first 50 strides compared with the FES condition as shown in Fig. 5. This suggests that the subjects tried to move their ankle.

Although the FES and the vL-FES condition are considered to be effective for gait training, muscle fatigue was caused within a few tens of minutes and the voluntary effort was not sustained for long duration. Fatigue resistance of muscle is considered to be improved by using FES frequently. When gait rehabilitation training under the vL-FES is applied to several times of 10 m walking or to walking for a short time, it can be effective. To apply the developed portable rehabilitation system to home rehabilitation, improving FES movements after muscle fatigue and inducing voluntary effort for long time have to be studied. It is also desired to show rehabilitation effects of gait relearning using the system increasing the number of subjects.

6. Conclusion: A prototype of portable FES rehabilitation system using the foot drop correction was developed for rehabilitation training of relearning gait of healthy subjects. Electrical stimulation timing for the foot drop correction or providing gait information was detected by the inertial sensor attached on the foot, which was determined based on gait event timings and muscle activation timing during walking of healthy subjects. The toe off and the foot flat timings detected by the inertial sensor were suggested to be useful to determine the stimulation timing. Gait information can be measured with the developed system during gait training. The developed portable system was considered to be useful for gait rehabilitation under the FES and the vL-FES conditions through examinations with hemiplegic subjects. A way of FES rehabilitation after muscle fatigue and a method of applying the vL-FES in rehabilitation training have to be studied for practical use.

7. Acknowledgments: The authors thank Mr. N. Kuge, Mr. K. Murakami and Mr. H. Miura for supporting measurements with hemiplegic subjects.

8. Funding and declaration of interests: This work was supported in part by the Ministry of Education, Culture, Sports, Science and Technology of Japan under a Grant-in-Aid for Scientific Research (B). Conflict of interest: none declared.

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