The 2009 IEEE/RSJ International Conference on
Intelligent Robots and Systems
October 11-15, 2009 St. Louis, USA

A human interface for stride control on a wearable robot

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Abstract— A legged locomotor device for paraplegics have been attempted to improve their ADL and to prevent some complications. A stride control of the system based on the user's intension is important to coordinate the voluntary movements of the user and the assisted movements of the paralyzed legs. In this paper, we propose a human interface with a walker to control the stride length of a legged locomotor device. Assuming that a intended stride is equal to a distance of the preceding movement of the walker, we developed a human interface estimating the movement distance of the walker, where the distance is calculated by polynomial fitting for the acceleration of the movement. In this study, we examine the proposed human interface from the measurement experiments of gait movements, and report the following results: (1) estimation accuracy by polynomial fitting method, and (2) feasibility of the adjustment of stride length using the proposed method. These results suggest that the proposed human interface is effective to adjust the stride length of a legged locomotor device.

I. INTRODUCTION

Gait reconstruction systems for paraplegics have been attempted to improve their activities of daily lives and prevent some secondary complications [1]. Although functional electrical stimulation (FES), and hybrid systems of FES and a gait orthosis [2] were considered to reconstruct their leg motor function, they had some problem related to slow speed of walking [3], high energy cost [4], and difficulties of control [5]. Wearable robotic systems have recently been developed to assist the movement of disabilities, and they can accurately control paralyzed limb movements. In addition, the systems perform various movement patterns with many degrees of freedom.

A human interface system is an essential part of the robotic system to control the paralyzed limb based on the user's intension [6]. Especially, the human interface for gait assistance requires high manipulability because the user must operate multi-DOF system during the movement. In previous studies of wearable robotics, EMG signal is used for human interface to detect the user's intension [7], [8]. However, voluntary EMG signal can not be detected from leg muscles of paraplegics. In addition, involuntary muscle contraction occurs due to spasticities. It is difficult to assist the leg movement of paraplegics using the EMG signal of leg muscles.

This study was supported by The New Energy and Industrial Technology Development Organization (NEDO), Grant-in-Aid for Scientific Research (B) 18360202 from JSPS and for Young Scientists (B) 20760165 from MEXT, and Electro-Mechanic Technology Advancing Foundation.

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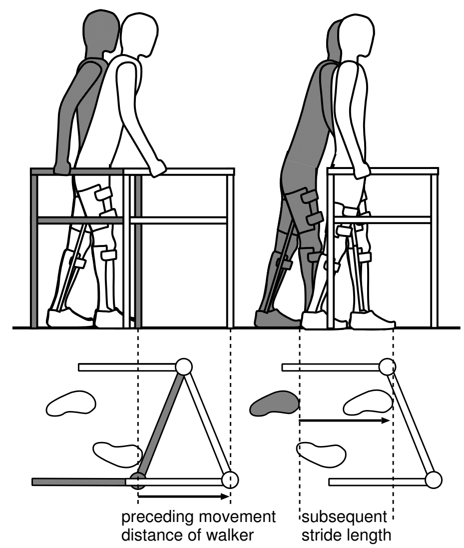


Fig. 1. A gait pattern with a walker. A user moves his arm with the walker, and then the opposite leg is moved by WPAL. In a cyclic gait, the preceding movement distance of the walker should be equal to the subsequent stride length.

To maintain upright posture during paraplegic walking with an orthosis, arm-support with a walker or crutches is required [9]. In this paper, we propose a human interface to control a stride length based on the relationship between a voluntary movement with a walker and an assisted leg movement. Fig. 1 shows a typical movement sequence during paraplegic walking with a walker. A paraplegic move their arms before leg movement. Assuming a cyclic gait movement, the distance of the preceding arm movement is equal to the stride length of the subsequent leg movement. Therefore, the intended stride length can be predicted by measurement of the distance of the preceding arm movement. In our human interface, triaxial angle-acceleration sensors are attached on the walker. The movement distance of the walker is estimated from measured acceleration data. The gait pattern of a wearable robot is determined so that the stride length is equal to the estimated distance of the preceding walker movement.

Aiming to evaluate the performance of our human interface, we conducted two experiments in the following viewpoint: (1) Estimation accuracy of walker movement distance using acceleration data, (2) feasibility of our system during walking.

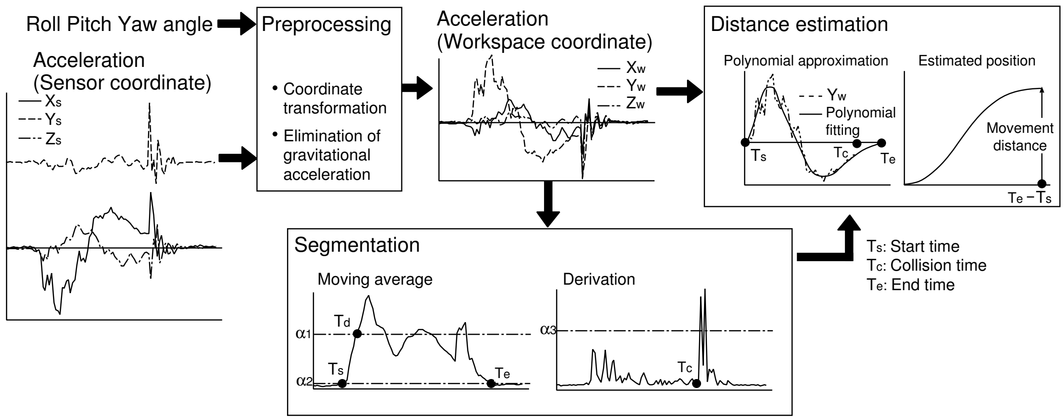
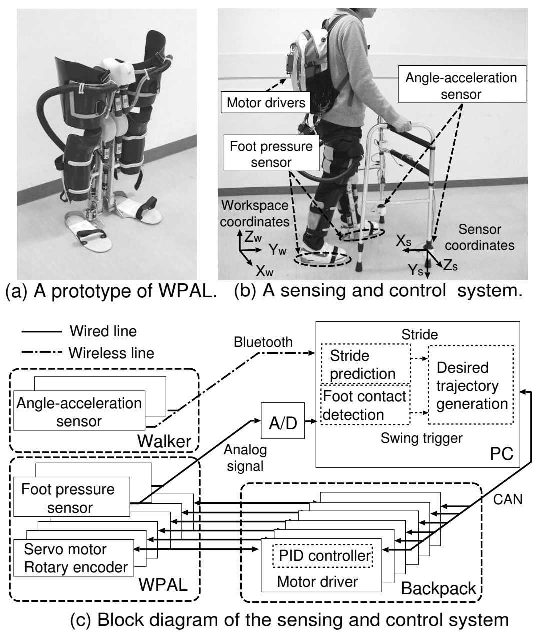


Fig. 3. A signal processing flow of the distance estimation using acceleration of a walker movement.



(a) A prototype of WPAL. (b) A sensing and control system.

(c) Block diagram of the sensing and control system

Fig. 2. A prototype system of WPAL (Wearable Power-Assist Locomotor). (a) shows the robotic equipment with 6 actuated joints consisting hip, knee and ankle joints. (b) shows the sensor and control system for locomotion with a walker. Two accelerometers are attached on the front frames of the walker, and foot pressure sensors are attached on the plantar parts. (c) shows the block diagram of the sensing and control system of the proposed interface.

II. ROBOTIC SYSTEM AND HUMAN INTERFACE

In this study, we developped a human interface for a wearable robot WPAL (Wearable Power-Assist Locomotor) shown in Fig. 2 [10]. WPAL is based on an unpowered gait orthosis 'Prime Walk' for paraplegics [11], and has six DC servo motors on hip, knee, and ankle joints to assist walking, stand-up and sitting down movements. The parts contacting the user's body is customize for individual.

As shown in Fig. 2(b), voluntary movements of a user are measured by angle-acceleration sensors and foot pressure sensors. The angle-acceleration sensors are attached on the front frame on the walker. Pressure sensors are attached to the foot sole of the WPAL. The control system is shown in Fig. 2(c). The main controller (PC) determines the desired trajectories of the actuators, and motor drivers control rotational angle of the actuators using PID feedback control. The angle, acceleration and foot pressure data are collected in the PC. The movement distance of the walker is estimated from acceleration data, and desired trajectories of the actuators are determined. The foot-floor contact information is provided by the foot pressure sensor. A leg swing movement is triggered by toe-off detection. In the following section, a estimation method of walker movement using acceleration data and generation of desired trajectories are described.

A. Movement distance estimation

Fig. 3 shows the signal processing of the acceleration data to estimate the movement distance. In the preprocessing block, the acceleration signal is transformed from the sensor-coordinates to the workspace coordinates and the gravitational acceleration is eliminated. A start time T_s and an end time T_e are detected from 5-point moving average, and a collision time of the walker and the floor is detected from the derivation in the segmentation block. Finally, a movement distance is estimated from movement time $T_e - T_s$ and acceleration data from T_s to T_c.

It is difficult to estimate a distance by double integral of acceleration data because of electrical and mechanical noise, filtering distortion, and bias drift [12]. We estimate a movement distance based on 'criteria of smoothness' known as an optimal criteria of human arm movements [13], [14]. Let a cost function be the square summation of n-th derivation of the hand position, the optimal hand trajectory is represented by

$$y(t) = \sum_{i=0}^{2n-1} a_i t^i, \tag{1}$$

where y indicates the position in progressional direction,

t indicates time, and a_i indicates a coefficient of i-th order term. The number of estimation parameters can be reduced using boundary conditions. The position, velocity, and acceleration of hand at the start time ($t = 0$) is equal to zero, and the velocity and acceleration at the end time ($t = t_f = T_e - T_s$) is also zero. The coefficients a_0, a_1 and a_2 can be zero from the boundary conditions of the start. From the boundary conditions of the end, the coefficients a_3 and a_4 can be represented as

$$a_3 = \frac{1}{3}\sum_{i=5}^{2n-1}(i^2 - 4i)t_f^{i-3}a_i \tag{2}$$

$$a_4 = -\frac{1}{4}\sum_{i=5}^{2n-1}(i^2 - 3i)t_f^{i-4}a_i. \tag{3}$$

The acceleration of hand position can be formulated from the boundary conditions.

$$\ddot{y}(t) = \sum_{i=5}^{2n-1} k_i(t)a_i \tag{4}$$

$$\begin{aligned} k_i(t) = {} & (t^{i-2} - 3t^2 t_f^{i-4} + 2t \times t_f^{i-3})i^2 \\ & +(-t^{i-2} + 9t^2 t_f^{i-4} - 8t \times t_f^{i-3})i. \end{aligned} \tag{5}$$

The parameter values of a_i ($i = 5, \cdots, 2n-1$) are determined by the least squared method minimizing error of the measured acceleration data from the start time to the collision time. The error function is

$$E = \frac{1}{2}\sum_{m=1}^{M}\{\hat{\ddot{y}}_w(t_m + T_s) - \ddot{y}_w(t_m)\}^2, \tag{6}$$

where M indicates the number of data of acceleration, and t_m represents time $m\Delta t$ (Δt is the sampling period). $\hat{\ddot{y}}_w(t)$ is progressional acceleration transformed to the workspace coordinates. The parameter values of polynomial function is determined by the following equations.

$$\boldsymbol{a} = \boldsymbol{A}^{-1}\boldsymbol{b} \tag{7}$$

$$\boldsymbol{A}_{ij} = \sum_{m=1}^{M} k_{i+4}(t_m)k_{j+4}(t_m)$$

$$\boldsymbol{b}_i = \sum_{m=1}^{M} \hat{\ddot{y}}_w(t_m + T_s)k_{i+4}(t_m).$$

Substituting the determined parameter $\boldsymbol{a}$ and movement time t_f to (1), the movement distance is

$$y_w(t_f) = \sum_{i=3}^{2n-1} a_i(t_f)^i. \tag{8}$$

B. Generation of desired trajectories

The desired trajectories of the joint angle are generated so that the stride length is equal to the estimated distance of walker movement. Fig. 4 shows the initial and terminal leg posture for leg swing movement, where S indicates desired stride length, θ_0^{hip} indicates initial angle of hip joint, and θ_f^{hip} indicate terminal angle. Assuming

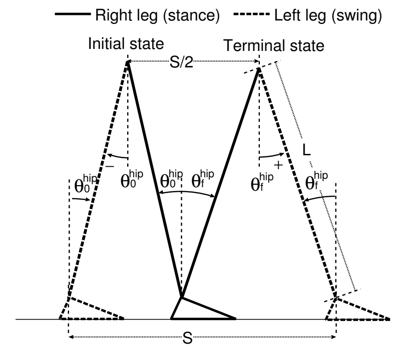


Fig. 4. Geometric relationship between the joint angle at toe-off (initial state) and that at heel strike (terminal state). A desired terminal state can be calculated from an initial state and a desired stride length.

that knee joints are maximum extension position and the foot sole of both legs are contact on the floor, the hip joint angle of terminal posture is

$$\theta_f^{hip} = \mathrm{Sin}^{-1}\left\{\frac{S}{2l} + \sin(\theta_0^{hip})\right\}, \tag{9}$$

where l indicates the length of leg. The foot joint angle of swing leg is plantarflexion position and that of stance leg is dorsiflexion position, and the size of these angle is equal to the hip joint angle. The pattern of joint trajectories are given from measured data of human gait, and the amplitude and offset are adjusted from the initial and terminal joint angle. Fig. 5 shows the joint angle patterns and stick picture when stride length is 0.4 m.

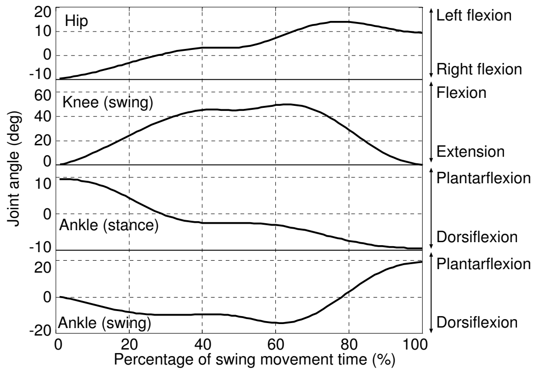
III. EXPERIMENTS

To examine the estimation accuracy for the polynomial fitting method, the walker movements during gait were measured in the experiment 1. We compared the estimation accuracy of the polynomial fitting method with a double integral method. In the experiment 2, the gait movements with WPAL and our interface system were measured. The feasibility of our system and correlation between the stride length and preceding arm movement distance are examined.

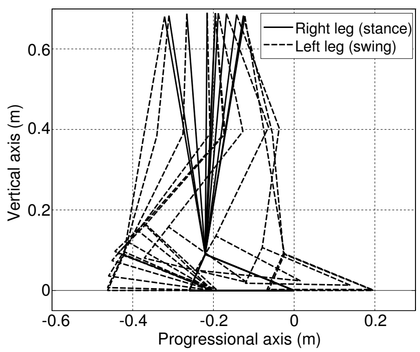
A. Experiment 1

Six subjects participated in this experiment. They had no disabilities of their motor function and neurological diseases. They were informed about the objectives and procedures of this experiment and gave written consent for their participation.

The experimental environment is shown in Fig. 6. The subjects instructed to walk in specified stride with a walker attached angle-acceleration sensors. Marks indicating the foot placement were attached on the walking path. Two LED markers were attached on the sensor and the positions were measured by 3-dimensional position measurement device (OPTOTRAK, Northern Digital Inc.). The stride length was set from 0.2 m to 0.55 m with interval 0.05 m.



(a) Joint angle



(b) Stick picture (S=0.4 m)

Fig. 5. A desired leg movement trajectory when the stride length is 0.4 m. (a) shows the time series of hip, knee, ankle joints, where the angle of the stance knee joint is fixed at maximum extension position. (b) shows a stick picutures for the gait pattern.

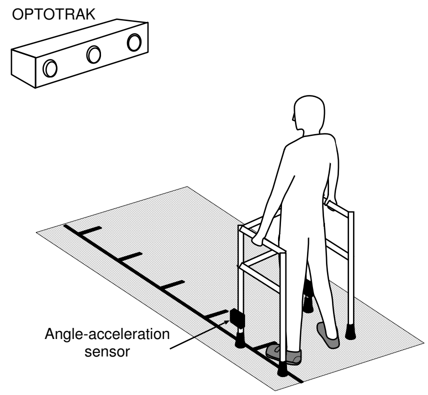
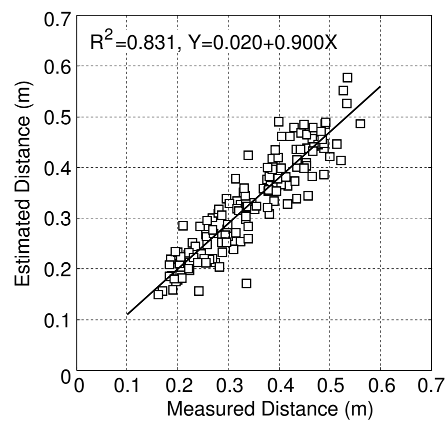
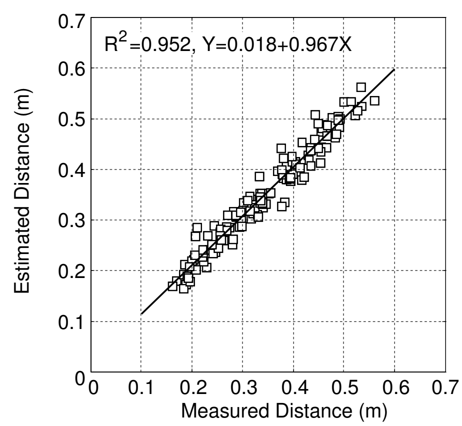


Fig. 6. An experimental setup for gait measurement with a walker.



(a) Double Integral method



(b) Polynomial fitting method

Fig. 7. Regression estimation of (a) double integral method and (b) polynomial fitting method for subject E.

The estimation accuracy is evaluated with linear regression analysis between measured and estimated distance of the walker during walking. To examine the effectiveness of our polynomial method, the results were compared with the results of the double integral method. From the performance of preliminary experiments, we applied the 7-th order polynomial function in proposed method.

B. Experiment 2

In this experiments, one normal subject who is one of the authors participated with custom-made WPAL for him. He put on the WPAL and instructed to walk with specified stride length. The positions of left and right toe and walker were measured by OPTOTRAK. The stride length was set from 0.2 m to 0.45 m with interval 0.05 m. In addition, the values of feedback gain were set high value so that the subject could not carry out voluntary leg movement. In the data analysis, the correlation between the stride length and preceding walker distance was examined.

IV. RESULTS

A. Experiment 1

Fig. 7 shows the regression results between measured and estimated distance of walker movements for a typical subject (subject E), where (a) shows the result for the double integral method and (b) shows that for the polynomial fitting method. The results of polynomial fitting shows higher linearity and lower variance than that of the double integral method. The regression lines shows that the polynomial fitting method is more accurate than the double integral method. Table I shows the regression

TABLE I

Results of regression analysis between estimated movement distance and measured distance of walker.

	Double integral method				Polynomial fitting method			
Subject	R^2	Slope	Intercept	Error (SD) [m]	R^2	Slope	Intercept	Error (SD) [m]
A	0.8990	0.9332	0.0109	0.0253 (0.0230)	0.9364	0.9934	-0.0096	0.0209 (0.0192)
B	0.9150	0.9314	0.0165	0.0199 (0.0181)	0.9553	1.0095	-0.0007	0.0159 (0.0118)
C	0.9255	0.9594	0.0044	0.0216 (0.0177)	0.9578	0.9707	0.0077	0.0147 (0.0132)
D	0.8145	0.9544	0.0024	0.0240 (0.0193)	0.9362	1.0496	-0.0026	0.0231 (0.0189)
E	0.8313	0.9003	0.0198	0.0340 (0.0275)	0.9521	0.9674	0.0176	0.0177 (0.0142)
F	0.7777	0.9613	-0.0034	0.0540 (0.0425)	0.9302	0.9804	0.0263	0.0302 (0.0257)

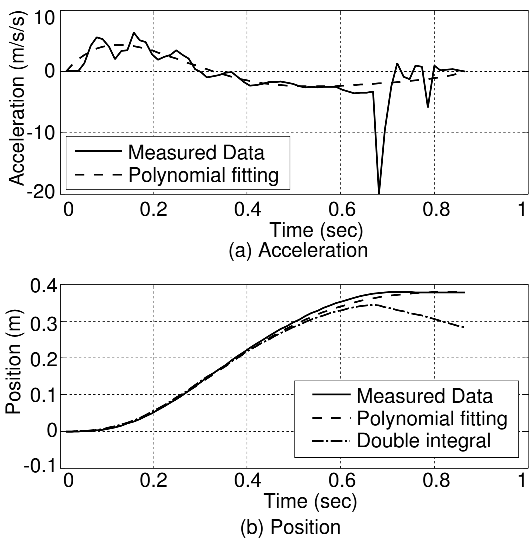


Fig. 8. Measured and estimated profiles of a typical movement of the walker for progressional direction. (a) A solid line and a dashed line indicate the acceleration measured by an inertial sensor and that approximated by the 7th order polynomial function, respectively. (b) A solid line indicates the position measured by OPTOTRAK, and a dashed line and a dash-dotted line show the positions estimated by the polynomial fitting method and the double integral method, respectively.

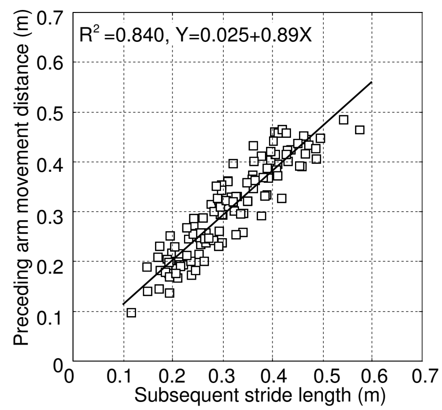


Fig. 9. Results of regression analysis between the preceding movement of the walker and subsequent stride length during the assisted walking with WPAL.

results of all subjects. Comparing the values of coefficient of determinant R^2 for the two method, The values of the polynomial fitting are larger than that of the double integral, and are higher than 0.9 for all subjects. In the double integral method, the values of R^2 show large variance among the subjects. The regression lines and error values show that the polynomial is more accurate than the double integral method. Fig. 8 shows the typical pattern of walker movement. The acceleration of polynomial fitting shows smooth for collision at t =0.64 sec. The positions of the double integral has large error after the collision while the position of polynomial fitting is good consistency with the measured position.

B. Experiment 2

Fig. 9 shows the results of regression analysis between the stride length and the movement distance of the walker. The results show good agreement of the movement distance of walker and stride length, while the value of R^2 is not as well as the results of Experiment 1. This would result from a gear-backlash of actuators and tracking errors in the PID controller. Fig. 10 shows time series of sensor outputs for robot-assisted gait with nine steps. A leg swing movement induced after a contralateral arm movement and an ipsilateral toe off. These sequential movements were repeated during walking. Comparing the left acceleration pattern of the first and third movements, the amplitude of the first movement is less than that of the third movement. Then the amplitude of the first hip joint movement is less than that of the third movement. The first step is a transferring movment from standing to walking, and the stride should be in half of other swing movment. The subject moved the walker with small distance for small stride in the first step movement.

V. Discussion

In this paper, we propose a human interface to control a stride length according to the user's intension. Acceleration of a walker movement is used to estimate stride length of a subsequent leg swing movement, and a trigger of a swing movement is detected from data of foot pressure sensors. The user can control a stride length from a movement of the walker, and can trigger a swing movement from weight bearing toward his stance leg. Since these sensors are not attached on user's body, complicate preparations to use this system is not required.

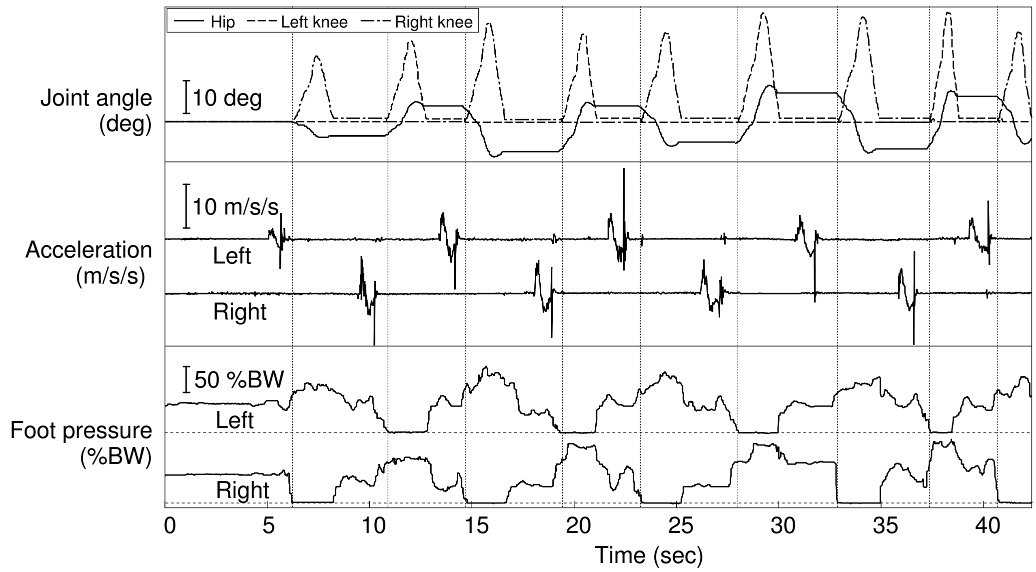


Fig. 10. Time series of sensor outputs during assisted walking with WPAL. Nine swing movements from a standing position are shown. In joint angle, the angles of hip, left knee and right knee are plotted. The acceleration profiles of left and right show those for the progressional direction of the walker movements. The profiles of right and left foot pressure sensors were normalized so that the mean value during a quiet standing is equal to 50 % for body weight.

The results of the experiment 1 showed that the polynomial fitting method is effective to estimate the movement distance. The evaluation experiment of proposed human interface demonstrated that it was feasible to control the stride length by adjusting the walker movement. These results suggests that the proposed human interface is effective for control of stride length of a wearable robotic system.

In future works, we will examine the effectiveness of the system for paraplegic patients. The patients, who can walk with an orthosis, might be possible to walk with our system because their motor skill of arm is as well as the normal people. However, postural and gait stability would be a limiting factor because a user cannot move his arm smoothly under an unstable posture. Modulation of gait pattern according to user's imparement and motor skill would be important for practical application.

VI. ACKNOWLEDGMENTS

The authors would like to thank professor Eiichi Saitoh and associated professor Yoshihiro Muraoka in Fujita Health University, and associated professor Naohiro Fukumura in Toyohashi University of Technology.

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