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Development of Third-Generation Intelligently Controllable Ankle-Foot Orthosis with Compact MR Fluid Brake

Takehito Kikuchi, Member, IEEE, Sosuke Tanida, Kikuko Otsuki, Takashi Yasuda, and Junji Furusho

Abstract— **Ankle–foot orthoses (AFOs) are orthotic devices that support the movement of the ankles of disabled people, for example, those suffering from hemiplegia or peroneal nerve palsy. We have developed an intelligently controllable AFO (i-AFO) in which the ankle torque is controlled by a compact magnetorheological fluid brake. Gait-control tests with the i-AFO were performed for a patient with flaccid paralysis of the ankles, who has difficulty in voluntary movement of the peripheral part of the inferior limb, and physical limitations on his ankles. By using the i-AFO, his gait control was improved by prevention of drop foot in the swing phase and by forward promotion in the stance phase.**

Index Terms— **Ankle foot orthosis, Compact brake, Gait analysis, Gait control, Magnetorheological fluid**

I. Introduction

LOCOMOTION is an important skill for the activity of daily living for humans. Gait training is therefore given a high priority in rehabilitative training.

Normal gait is cyclic and can be characterized by the timing of the foot contact with the ground. An entire sequence of functions by one limb is known as the gait cycle (Fig. 1) [1]–[3]. Each gait cycle has two basic components: the "stance phase", during which the foot is in contact with the ground, and the "swing phase", during which the foot is in the air for the purpose of limb advancement. The swing phase can be further divided into three functional subphases: initial swing, mid swing, and terminal swing. In the same manner, the stance phase can be divided into five functional subphases: initial contact, loading response, midstance, terminal stance, and preswing [1], [4], [5]. In normal gait, the initial contact is heel-contact (or "heel-strike"). In the initial swing (or "toe-off"), normal subjects are able to maintain an appropriate clearance between the tip of the toe and the ground to prevent any inappropriate interactions with the ground.

Patients who have a dysfunction of the ankles, for example those suffering from polio or peroneal nerve palsy, have difficulties in controlling their ankle movements. This causes "drop foot" or a lack of dorsal flexion of the ankle during the swing phase. In many cases, such patients are unable to prevent themselves from catching their toes on the ground and stumbling, even when taking small steps. Additionally, these patients tend to incline their bodies more than do healthy persons because of the motion required to prevent stumbling. This causes an undesirable loss of energy in walking.

Orthoses are devices that are attached or applied to the external surface of the human body to improve functions or to restrict, enforce, or support a body segment [1], [6]. Lower-limb orthoses can be used to improve the gait functions of patients, and to assist ankle function, ankle–foot orthoses (AFOs) are often used to restrict involuntary plantar flexion.

Powered AFOs have been the focus of several recent studies. A number of types of powered AFO have been reported [7], and these use several types of actuator, e.g. a pneumatic actuation system [8], a ball screw drive system [9], or a series elastic actuator [10]. However, the development of such powered AFO devices is more challenging than that of powered prostheses, because the dynamics of the original leg and foot (inertia, viscosity, elasticity, and voluntary/involuntary forces from spastic limbs) must be considered. Additionally, there are more-severe demands on weight saving.

Passive orthotic devices have been suggested as an alternative design concept for a controllable orthosis. In particular, drop foot can be controlled by using a passive device alone. Passive controllable AFOs also have considerable advantages in terms of cost, safety, and miniaturization. Berkelman et al. [11] developed a completely passive orthosis with only a parallel linkage mechanism; however this device requires extensive adjustment to the needs of the individual user. Farris et al. [12] suggested a joint-coupled orthosis that uses wafer-disk friction brakes to control the torque at hip joints and knee joints. However, the response time of their brake is about 150 ms, which is insufficient to assist in the rapid dynamics of the human gait; for example, the loading response (Fig. 1) normally ends after 10% of a gait cycle or about 100 ms.

We have designed a passive controllable AFO with a compact magnetorheological fluid brake (CMRFB) [13] for dynamic control of the gait. The CMRFB is capable of a rapid and stable torque response. Its response time is faster than that of other types of conventional brake, e.g. powder brakes. In a previous report [14], we described several types of intelligently controllable AFO (i-AFO) each of which use a CMRFB as a torque generator. Here, we

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Takehito Kikuchi is with the Graduate School of Science and Engineering,, Yamagata University, 4-3-16 Jonan, Yonezawa 992-8510 Japan (corresponding author, phone & fax: +81-238-26-3892; e-mail: t_kikuchi@yz.yamagata-u.ac.jp).

Kikuko Otsuki and Junji Furusho are with the Graduate School of Engineering, Osaka University, 2-1 Yamadaoka, Suita, Osaka 565-0871, Japan

Sosuke Tanida and Takashi Yasuda are with the Shiga School of Medical Technology, 967 Kitasaka, Higashiomi, Shiga 527-0145, Japan.

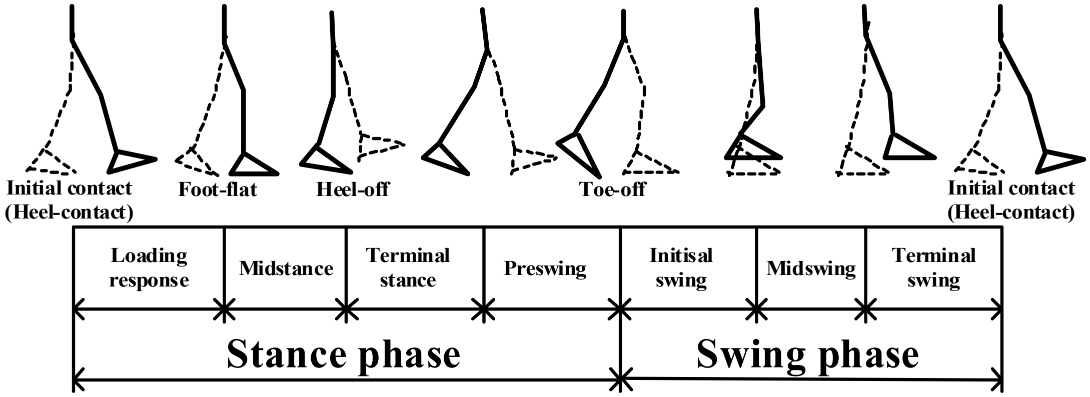


Fig. 1. Normal gait cycle

describe the development of a new CMRFB that has a lower mass than the previous brakes, and a new i-AFO that uses this brake. This is the third generation of our i-AFOs. Additionally, we revised the control method to improve the intelligence of the i-AFO. We tested the new i-AFO on a patient with post-Guillan–Barre syndrome, and we conducted gait-control tests under several different sets of experimental conditions to determine its usefulness.

II. Intelligently Controllable Ankle Foot Orthosis

A. Overview

Fig. 2 shows the newly developed third-generation i-AFO. The specifications for this device are listed in Table I. This orthotic device was developed for use in gait rehabilitation or to assist walking in daily life for patients who have motor dysfunctions in their ankles. Target users are patients who can walk stably with a plastic AFO. Major concept of the i-AFO is the same as that of the previous system [14], however, we improved the sensor system and control method as the next sections.

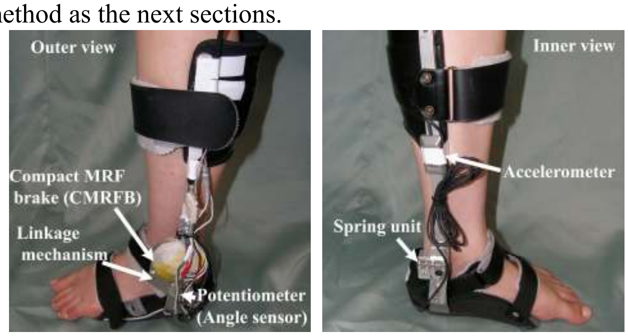


Fig. 2. Third-generation i-AFO with 5 Nm-class CMRFB

TABLE I
SPECIFICATION OF THE THIRD-GENERATION I-AFO

Maximum braking torque [Nm]	10
Mass [g]	990
Movable angle [deg]	–45 to +45

As shown in the picture, the AFO system consists of the following components:

(1) a compact MRF brake (CMRFB) for ankle torque control,
(2) a linkage mechanism to amplify the torque of the CMRFC,
(3) a magnetic rotary potentiometer to measure the angle at the ankle,
(4) an accelerometer to detect initial contact, and
(5) a spring unit to assist the brake torque at the initial contact.

The total mass of the i-AFO is 990 g. This is slightly heavier than that of commonly used metal-type AFOs. Each component is described below.

B. Compact Magnetorheological fluid brake

We developed a CMRFB for installation in the ankle joint of the i-AFO to control the ankle torque. We set a target of 5 Nm for the maximum torque of this brake, because in a previous study [14] we found that the torque required to prevent abnormal plantar flexion in the swing phase is less than 10 Nm in many cases, and we can amplify the brake torque by using a linkage mechanism, as described below (see Section II C).

On the basis of the multilayer structure discussed in a previous study [13], we developed a 5 Nm-class CMRFB. Figs. 3 and 4 show a picture and a cross-sectional view of the CMRFB, respectively. Table II lists the specifications for the 5 Nm-class CMRFB. The multilayered disks are fixed with accurate gaps of 50 μm. These gaps are completely filled with the MRF (140CG, Lord Corp.) [15]. A piston mechanism is used to prevent leakage of the MRF as a result of thermal expansion. The core of the electromagnet is made of silicon steel, and the casing is made of aluminum. The diameter of the magnetic wire is 0.2 mm.

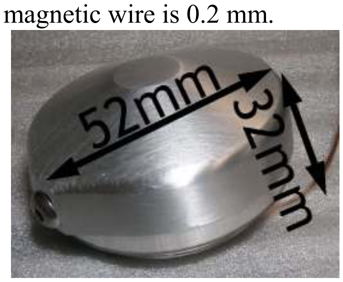


Fig. 3. 5 Nm-Class CMRFB

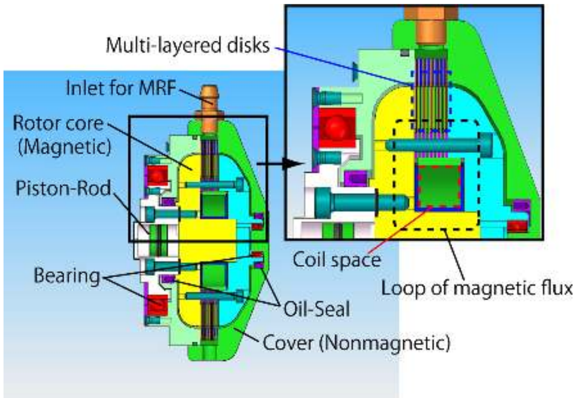


Fig. 4. Cross-section of the 5 Nm-class CMRFB

TABLE II
SPECIFICATIONS FOR THE 5 NM-CLASS CMRFB

Total thickness [mm]	32
Outer diameter [mm]	52
Diameter of disks [mm]	40
Number of disks	9 input + 8 output
Number of MRF layers	18
Gap size of MRF layers [μm]	50
Number of turns in the coil	191
Idling torque [Nm]	0.15
Max. torque at 1 A [Nm]	>5.0
Time constant of torque [ms]	10
Mass [g]	237

Time constant of torque of this device is about 10ms which is greatly faster than any other conventional brakes. Thanks to this rapid response, the CMRFC can control rapid dynamics in the initial stance of human. On the other hand, one of the disadvantages of this device is its high energy consumption. If electric current is applied constantly, the CMRFC consumes 3~6W. In the previous report, we apply constant electric current to the CMRFC in order to maintain the dorsal flexion of ankle in the swing phase, which requires high energy. In this paper, we suggest using a velocity feedback in the swing phase and it helps this system to reduce the energy consumption.

C. *Linkage mechanism for torque amplification*

As shown in Table II, the CMRFB can only generate about 5Nm of torque. To enhance the ankle torque of the i-AFO, we used the linkage mechanism shown in Fig. 5. As shown in the left-hand diagram, the CMRFB is located a little above (~40 mm) the ankle joint and is connected by Link 1. Link 2 is an output link that is connected to the foot part of the i-AFO.

The ratio of the length of Link 1 to that of Link 2 is 20:35, as shown in the right-hand figure. The angle limitation is –45° (dorsal flexion) to +45° (plantar flexion). The torque amplification ratio of the braking torque therefore changes from 1.7 to 2.0, depending on the angle of the ankle. We can measure the angle by means of a rotational potentiometer and calculate the amplification ratio in real time.

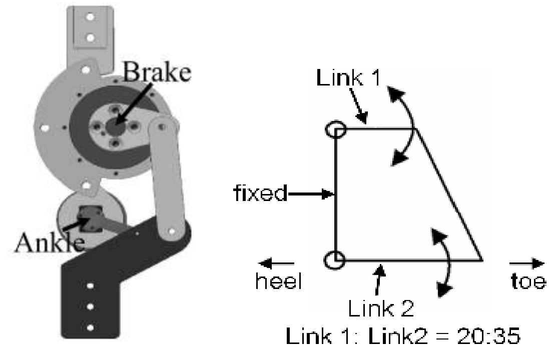


Fig. 5. Linkage Mechanism

D. *Spring unit*

The maximum controllable torque in the i-AFO, produced by the CMRFB and the linkage mechanism described above, is about 10 Nm. This is sufficient to prevent abnormal drop foot in the swing phase, but insufficient to control plantar flexion in the loading response (Fig. 1). To assist the ankle torque in plantar flexion in the loading response, we installed a spring unit on the ankle joint of the inner side of the i-AFO. Fig. 6 is a schematic representation of the spring unit. A lever arm pushes the spring during plantar flexion of the ankle only. We can adjust the spring constant by changing the spring. In this case, we installed a spring with a spring constant of 30.6 Nm/rad.

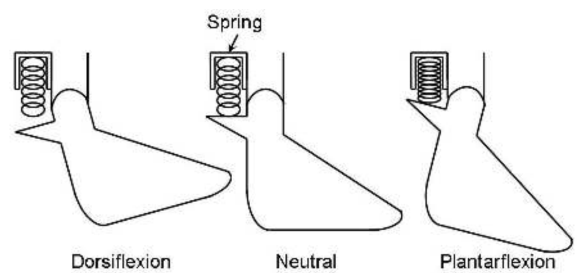


Fig. 6. Schematic of the spring unit

E. *Sensing system*

Until the previous version of i-AFO [14], we had detected the initial contact (heel contact) by means of foot switches attached on the bottom of the sole. However, the use of foot switches has some disadvantages, including the following.

(1) They are very sensitive to the position and shape of the sole and to the environment.
(2) It is difficult to ensure durability of the switches.
(3) There is a delay in the reaction of the sensors.

To eliminate these disadvantages, we removed the foot switches from the sole and attached an accelerometer (MA3-04Ac-RDB; maximum acceleration: 4G; MicroStone Co. Ltd., Japan) on the metal brace to permit detection of contact.

Additionally, the ankle angle is measured by a rotary potentiometer (HSM12, measurable angle: 90°, Sakae

Tsushin Kogyo Co. Ltd., Japan) attached to the ankle joint.

III. CONTROL OF THE I-AFO

A. Purpose

The main purposes of control are as follows:

(A) to prevent slap foot and knee buckling at initial contact, and to advance a smooth gait; and

(B) to prevent drop foot in the swing phase and to ensure sufficient clearance above the ground.

Conventional plastic AFOs are used mainly for purpose (B). Some types of AFO have hydraulic dampers or friction brakes for purpose (A). However, they do not have a function that adjusts the damping or frictional parameters automatically.

B. Classification of gait states

For convenience in designing the control method, we renamed the phases of the gait cycle (Fig. 7) as follows:

State 1: from the initial contact (IC) to the foot flat (FF),
State 2: from the FF to the heel off (HO), and
State 3: from the HO to the IC.

We tried to realize purpose (A) (see Section III A) in State 1, and purpose (B) in State 3.

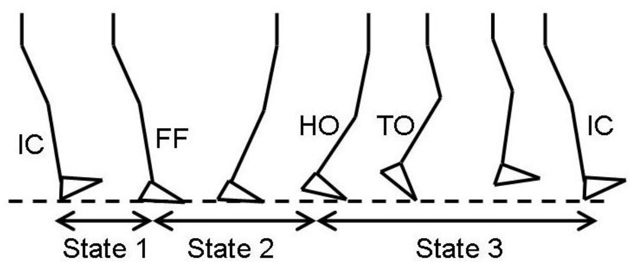


Fig. 7. Classification of gait states

C. Determination of gait states

Fig. 8 shows a flow diagram of the control method. The gait states are those described above in Section III B. In the flow diagram, the gait state shifts cyclically from State 1 through State 2 to State 3. Judgments 1, 2, and 3 are used to decide when the gait state shifts to the next stage. Each state continues while the result given by the corresponding judgment is "NO".

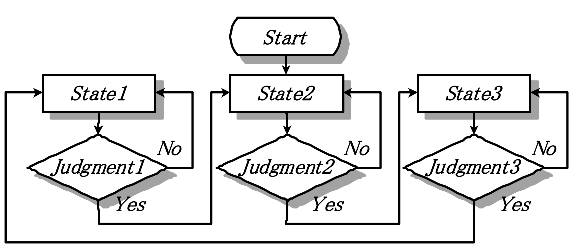


Fig. 8. Control flow

Subjects start gait experiments from an upright state. Therefore, the control starts from State 2. Each judgment is explained below. The control method for the brake will be explained later in Section III D.

Judgment 1: In this phase, the i-AFO identifies the start of the State 2 or FF. In this judgment, we used only angle information from the angle sensor. In State 1, after the IC, if the rotational direction of the ankle joint changes from dorsal flexion to plantar flexion, the control flow passes Judgment 1.

Fig. 9 shows the ankle angle and the state transition of a healthy person with the i-AFO. The solid line represents the ankle angle. The dashed line represents the state transition. As show in this figure, at FF, the angle direction changes from a negative direction (plantar flexion) to a positive direction (dorsal flexion).

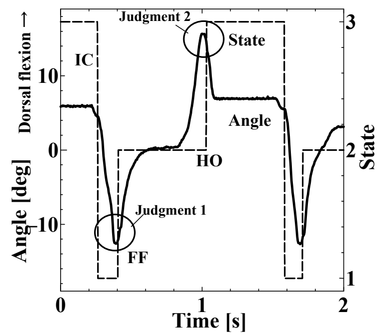


Fig. 9. Ankle angle and state transition

Judgment 2: In this phase, the i-AFO determines the start of State 3 or HO. In this judgment too, we use only the angle information from the angle sensor. In State 2, the ankle angle continues to rotate toward dorsal flexion. If the rotational direction changes from dorsal flexion to plantar flexion and if this change exceeds a selected constant angle ($\Delta\theta$ [deg]), the control flow passes Judgment 2. The value of $\Delta\theta$ is set to prevent misjudgments, and depends on the individual's gait. In this case, we set the value of $\Delta\theta$ to be 0.4°.

As shown in Fig. 9, at the HO, the angle direction changes from a positive direction (dorsal flexion) to a negative direction (plantar flexion). In addition, the angle remains constant in State 3, because the i-AFO controls the angle of the ankle. We will explain this in the next section.

Judgment 3: In this phase, the i-AFO determines the start of State 1 or IC. In this judgment, we made use of the acceleration in the vertical direction. If the accelerometer measures an acceleration of more than a threshold vale (α [m/s^2]), and if the duration from the previous judgment exceeds 75% of a gait cycle (t_{cycle} [s]), the control flow passes Judgment 3. The values of α and t_{cycle} were set to prevent misjudgments, and depended on the individual gait. In this case, we set α to 15 m/s^2. The value of t_{cycle} is

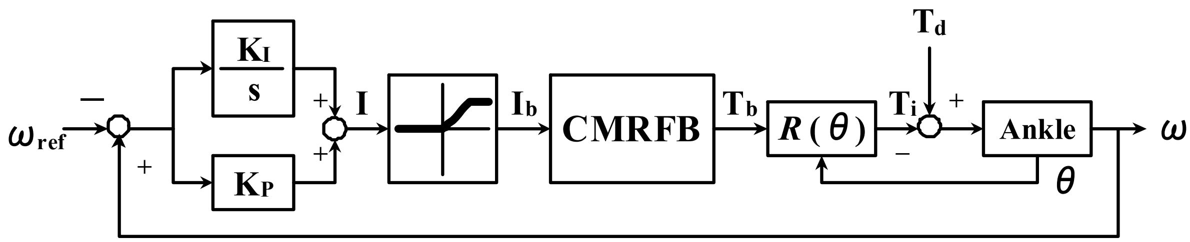


Fig. 11. Block diagram for velocity control

automatically calculated by the control system.

Fig. 10 shows the acceleration in the vertical direction and the state transition for a healthy person using the i-AFO. The solid line represents the acceleration, and the dashed line represents the state transition. As show in this figure, we observe large and rapid changes in acceleration at IC.

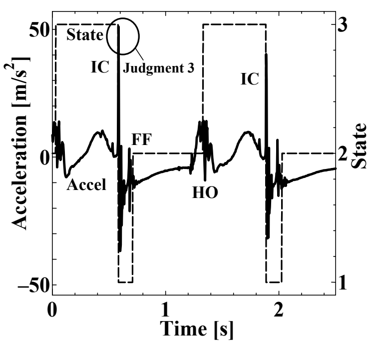


Fig. 10. Acceleration and state transition

D. *Torque control method*

In this section, the torque-control methods for each state will be explained.

State 1: The purpose of torque control in this state is purpose (A) mentioned above (see Section III A). The i-AFO (or the CMRFB) generates a braking torque to make the plantar flexion of the ankle joint from the moment of IC. In a previous report [14], we describe our attempts to control the ankle torque by means of a feed-forward model; however, it was difficult to achieve stable control because of disturbances, such as external forces or friction within the system. We therefore implemented an angular velocity control with a feedback model.

Fig. 11 shows a block diagram of the control system for State 1. The positive direction of rotation is defined as the direction of plantar flexion of the ankle. In this model, ω_{ref} [rad/s] is a reference angular velocity and the input for this system. We used a proportional integral (PI) controller as the controller. K_I [A/rad] is the integral gain and K_P [A*s/rad] is the proportional gain. In this case, we set K_I to be 0.001 A/rad and K_P to 0.3 A*s/rad.

The controlled object is the CMRFB installed on the i-AFO. The CMRFB generates a braking torque T_b [Nm] that depends on the input current I_b [A]. However, the brake cannot generate a torque in the direction of acceleration. Therefore, if the value of I_b calculated by the PI controller is negative, it is set to zero. Additionally, if the calculated value of I_b exceeds the maximum current (1.0 A), it is set to 1.0 A. The braking torque, T_b is amplified to T_i [Nm] by the linkage mechanism. The torque-amplification ratio is defined as R in this figure, and this value changes depending on the angle. We therefore have to calculate R in real time. T_d [Nm] is the disturbance torque from the environment; this consists mainly of the external torque when the heel hits the ground and the internal torque from the user. In our control system, the CMRFB generates a braking torque only when the real angular velocity ω [rad/s] exceeds ω_{ref}.

State 2: In this state, the i-AFO does not apply any torque; this allows smooth rotation of the ankle.

State 3: The purpose of torque control in this state is purpose (B) mentioned above (Section III A). The control method is same as that in State 1, but the reference angular velocity is set to zero to prevent drop foot.

IV. GAIT EXPERIMENT

We tested this i-AFO on a patient with post-Guillan–Barre syndrome, and we conducted gait-control tests under several experimental conditions to determine the usefulness of the device.

A. *Subject*

The subject was a male patient with post-Guillain–Barre syndrome (age: 34 years; height: 183.0 cm; mass: 83.1 kg). The subject has difficulty in voluntary movements of the peripheral part of the lower limb, especially the ankle joints and toes of both legs. He shows drop foot in walking. In addition, he shows considerable atrophy of the disused muscles. He usually attaches a plastic AFO to support the ankle function. The passive range of movement (ROM) of the ankle joint of the plantar flexion was 45°, and that of the dorsal flexion was 0°. He therefore had particular restrictions in dorsal flexion.

We obtained the standard form of informed consent, and we managed personal information strictly according to ethical guidelines.

B. *Method*

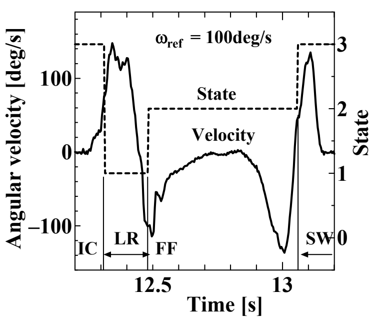
We examined two conditions in the gait tests. The i-AFO was used to controls the braking torque in State 1 with reference velocities of 100 and 200 degrees per second, respectively.

Before the experiments, the subject performed ambulatory exercises. In the experiments, the gait cycle was controlled by means of a metronome with a period of 1.3 s. In the preliminary tests, the subject declared that the optimal value for the reference velocity in State 1 was 200 degrees per second.

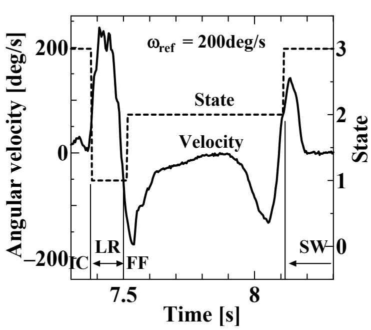
C. Experimental results and discussion

Fig. 12 shows the angular velocities in the experiment under the two conditions. The solid lines represent angular velocities of the ankle; the dashed lines represent the state transition of the gait. The i-AFO properly determined the gait states with information from the accelerometer and the rotational sensor. During the LR, the i-AFO controlled the angular velocity of the ankle joint to the reference values. Except for the initial of the swing phase, the i-AFO maintains zero velocity in the swing phase.

The user comfortably walks under the condition of 200deg/s, however, he feel discomfort in the LR under the condition of 100deg/s. The reference speed in State 1 should be set depending on the gait cycle. We will try to clarify the relationship between the gait cycle and this speed in the future.



(a) ω_{ref} = 100deg/s



(a) ω_{ref} = 200deg/s

Fig. 12. Angular velocities in the experiments

V. CONCLUSION

We have described the development of a third prototype for an intelligently controllable ankle–foot orthosis, and we have tested it in use under experimental conditions. The device includes a 5 Nm-class CMRFB as a torque generator. This brake has a good controllability (stable and rapid response of torque) and a high torque-to-weight ratio. By using this brake, we can develop a high-performance i-AFO. We also designed a controller for the i-AFO. In the new control system, we use an accelerometer and a rotary potentiometer to determine the state of the gait. To test the feasibility of the newly developed i-AFO, we conducted gait experiments on a patient with post-Guillan–Barre syndrome. The experiments showed that the i-AFO correctly controlled the angular velocity of the ankle joint by means of the velocity controller. The i-AFO worked as a velocity controller in the loading response and an angle limiter in the swing phase. These functions improved the abnormal gait of the subject. We can easily adjust some parameters of the i-AFO, so it should be possible to adjust the device to suite particular subjects or their recovery phases. Comparisons with other AFOs and healthy gait are future tasks.

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