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Ankle-Foot-Orthosis Control in Inclinations and Stairs

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***Abstract*—A control procedure is proposed for an ankle-foot-orthosis (AFO) for different gait situations, such as inclinations and stairs. This paper presents a novel AFO control of the ankle angle. A magneto-rheological damper was used to achieve ankle damping during foot down and locking at swing, thereby avoiding foot slap as well as foot drop.**

The controller used feedback from the ankle angle only. Still it was capable of not only adjusting damping within a gait step but also changing control behavior depending on level walking, ascending and descending stairs. As a consequence, toe strike was possible in stair gait as opposed to heel strike in level walking. Tests verified the expected behavior in stair gait and in level walking where gait speed and ground inclinations varied. The self-adjusted AFO is believed to improve gait comfort in slopes and stairs.

I. Introduction

Ankle-Foot-Orthoses (AFO) are today commonly used in helping people with reduced possibility of controlling their own foot ankle. The main purpose is to support and align the foot and ankle; suppress overpowering and to assist weak or paralyzed muscles. The designs are typically made of some plastic material keeping the ankle fixed and which has been found to provided a better foot prepositioning at heel strike, delayed heel rise and helped at toe clearance [11]. Existing commercial designs mainly handle foot drop but give no assistance to avoid foot-slap at foot down [12]. Miyazaki *et al.* found on hemiplegic users that the dorsi flexion muscle moment was weak or inactive at heel down making the orthotic aid important [15]. It was also noticed that the ankle-muscle moment was much larger than the orthotic moment during the rest of the stance phase. They concluded therefore that the "AFO only played a small role in assisting the plantarflexors". Here we use the term *dorsi flexion* when describing ankle angles between toe and calf which are less than 90 degrees and *plantarflexion* larger angles. In cases where only small dorsal flexion movement is needed a less stiffer AFO can be used giving a larger dorsi flexion movability making it less sensitive to ground variations [19].

Reports of active control in orthoses can be traced back to the 1960s. But the size and price of technology often limited the performance (see e.g. [4]). Large attention has been made to add force or power to the bio-mechanical system as an active powered ankle. The motivation lies both in the area of assisting handicapped people and as in lifting aid for able bodied i.e. exoskeletons (see e.g. [10], [9]).

An ideal AFO would follow a similar trajectory as a sound ankle. To control this continuously with an electric or pneumatic actuator could be cost inefficient i.e. power consuming and bulky. Recently several approaches have been suggested where the ankle stiffness can be varied by a spring in serial with a DC-motor [17] to avoid foot slap and foot drop. With force transducers and a rotary potentiometer gait phases were detected by comparing with thresholds. Hitt *et al.* proposed an actuator control which followed an individual ankle angle trajectory where the spring allowed the motion to differ [8]. The system was continuously updated for gait speed variations by a variable time-base of the trajectory. Instead Blaya and Herr showed that foot slap could be avoided in early stance and foot drop in swing by controlled stiffness. During the contact phase the stiffness was set to zero to avoid obstructing the plantarflexion movement [2]. Yamamoto *et al.* developed an ankle with cam-mechanism coupled to an oil damper [20]. The ankle had a resistive torque in only the plantar flexion rotational direction. This was shown to reduce the rapid plantarflexion at heel strike to foot flat. Furusho *et al.* [5] have developed an orthosis which avoids foot drop during foot down and at toe-off. A MR-brake minimized ankle motion from toe-off to foot flat where the damping was slowly reduced from heel strike to get a soft foot down movement. At weight acceptance was the orthotic ankle torque set to zero. With a force sensor on the foot sole, bending moment sensor on the shank and angle sensor a redundant system could detect gait phases. Initial test on a paralyzed polio person showed promising results.

The previously proposed active AFOs were designed for continuous gait on level ground. The controllers included used angle thresholds which may vary with ground circumstances and the purely mechanical solution could reduce the plantarflexion e.g. possibility for weight acceptance at stair ascending. This paper presents a method to *how the ankle can be adjusted to ground variations*, such as changing inclinations and stair gaits.

II. Orthotic Gait

In continuous walking, during one gait cycle from heel strike to heel strike, the variations of a sound foot ankle angle (Fig. 1) can be summarized as: 1) an initial plantarflexion to get a smooth foot down to foot flat, 2) followed by an

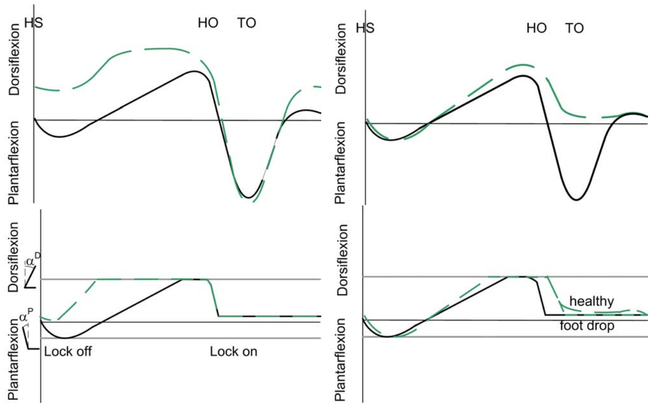

Fig. 1. Top; Sketch of a typical ankle angle variation during level walking for a sound foot (black) and walking in an inclination(green): ascending a hill (left) and descending (right) Bottom; principle from the desired controlled ankle foot orthosis. The mechanical device is limited in flexion to α^D and α^P. The cycle is from heel strike (HS) to heel-strike and also shows heel-off (HO) and toe-off (TO).

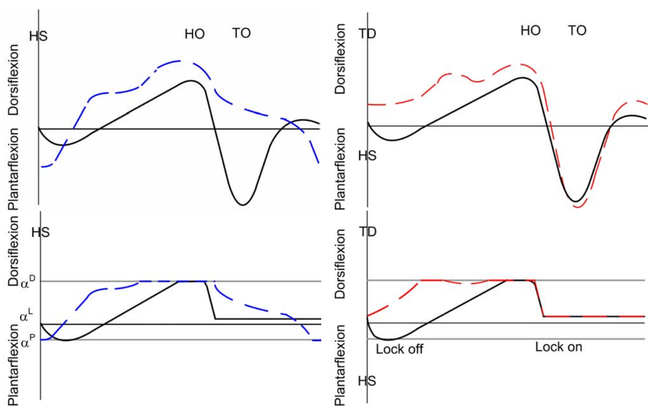

Fig. 2. Top; Sketch of a typical ankle angle variation at stair ascending (red/right), descending (blue/left) compared to level walking (black) and Bottom; principle from the desired controlled ankle foot orthosis. The cycle is from heel strike (HS) to heel-strike and also shows heel-off (HO) and toe-off (TO).

increasing dorsi flexion which ends at its maximum at heel off, 3) plantarflexion motion in foot lift and finally 4) during swing is the foot adjusted for avoiding foot drop. With a solid orthosis the possibilities of flexion are largely reduced. Maximum flexion is typical only a few degrees and comes from bending the material. The size of ankle flexion also increases with gait speed.

Using a hinged orthosis the ankle is allowed to articulate but generally without support for foot slap and drop. These are commonly designed with a mechanically limitation of maximum dorsi flexion and plantar flexion prventing the foot ankle from collapsing [12].

In up hill walking a sound ankle adjusts to ground inclination [14]. At stance the ankle angle is biased to a larger dorsi flexion (Fig. 1). The bias is proportional to the degree of inclination. Walking downhill the adaptation is not so significant in the ankle, but rather in the knee [7].

A. *Stair Ascending*

It has also been shown (see e.g. [1], [6]) that the ankle angle largely differs in stair gait compared to level walking (Fig.2). During weight acceptance the foot dorsiflexes at foot down and continuous to stay so in preparation for the opposite foot to be lifted. Foot down is not always initiated by heel strike, as in non stair cases, but more often with a front foot down causing this dorsi flexion. The pull-up motion is dominated by the knee while the ankle stays in a large dorsi flexion. Finally the body moves forward with a heel off upwards.

The ankle angle differs from horizontal gait mainly at the early stance phase and at the late swing. At the lift up to next staircase the edge is avoided by a small dorsi flexion and moving the knee backwards.

Conversely, a solid AFO limits the dorsi flexion at foot down at the calf motion needed by *pull-up* and at body lift to next staircase. It has also been shown that children using AFO's compensate this narrowed ankle motion with pelvic motion retraction [16].

B. *Stair Descending*

Also in down stair gait the ankle angle differs from horizontal gait in the swing phase when moving the limbs down [1], [13] (Fig. 2). In the early weight acceptance the toes are put down before the heel. During the foot lowering into the stair the weight is transferred from the opposite foot and the body motion continues forward. The body is lowered to meet the next staircase with the opposite foot. This also causes the largest difference in stance compared to level walking since a large dorsi flexion is needed to move the body both downwards and forwards in front of the foot.

A solid AFO would limit the initial plantar flexion preventing the foot to be put down with the toes first. Finally in the controlled lowering the solid design is unable to adapt to the large, 30-40 degree, dorsi flexion at heel lift [18]. Wearing a solid ankle AFO this has been shown to be compensated by pelvic and trunk [16]. During swing the effect of the AFO is not significant.

III. Active Control with Damper

We propose a hinged AFO with a controllable ankle flexion where the articulated ankle enables motion in the sagittal plane. The main objective of this design was to

- enable soft foot down,
- hinder foot drop during swing.

Previously described solutions showed that soft down for foot slap reduction was possible by adding a resistive force in the dorsal direction. While the foot drop could be prevented by keeping the angle constant during swing. This gives the desirable ankle angle behavior as shown in Fig. 1. In this case, the mechanical solution minimizes the risk of ankle collapsing by limiting the plantarflexion to α^P and dorsi flexion to α^D and the angle during swing is kept constant at α^L. For use in daily situations it also has to be less sensitive for ground variations. Thus it has to

- enable large dorsi flexions during stance phase (Fig. 1,2),

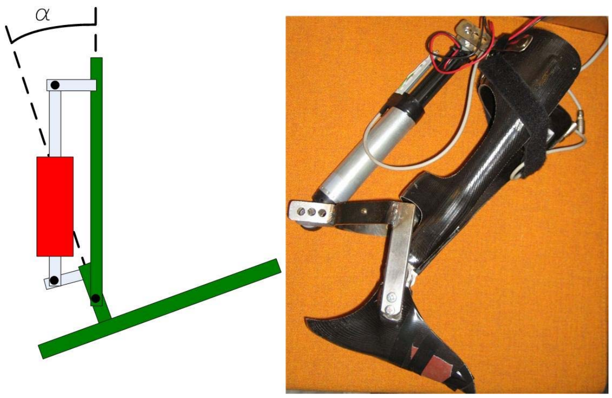

Fig. 3. Active orthosis prototype. Left: Sketch of working principle. With a linkage system is the ankle angle α adjustable. Varying the viscosity of the MR-damper (red) the ankle angle damping is adjustable or is kept fixed. Right: Test equipment.

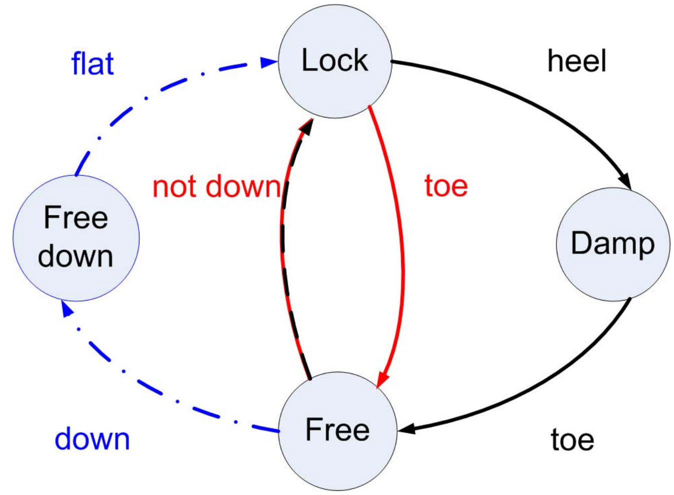

Fig. 4. Control states during three different walking conditions: Level(black), stair ascending(red) and stair descending (blue). Switching between states occur when the foot ground contact is initiated with the *heel*:$\Delta\alpha < 0$ or *toe*:$\Delta\alpha > 0$. Whether the ankle is to be locked is decided by identification direction being *down* or *not down* or when the gait is back on *flat* ground.

- allow plantar flexion during swing in stair descending.

From the latter demand the ankle includes the option of being unlocked during swing (Fig.2).

A. Actuation System

The actuator motion principle is to lock the orthoses during phases when the wearers muscle force is weak. A linear damper in a linkage system, as shown in Fig.3, determines the resistive ankle torque of the AFO. The rotation between locking positions is achieved by the wearers own motion.

The damper is of *magneto rheological*(MR) type [3] that is filled with a fluid which viscosity, thus damping force, can be changed with an electrical current.

The ankle angle is measured with a linear resistor similar to a goniometer and measures the damper-link-arm length. By trigonometry is the ankle α in Fig.3 found. Measuring the ankle angle, as will be shown, makes it possible to estimate gait states although the complete foot not is put down as in e.g. stair walking or on uneven ground.

B. Control Principle

The objective of the controller is to adjust the damper current depending on gait phase to be damping, locking or very small damping. But the sequence of states depends on the particular gait situation; level walking, stair ascending or stair descending. Here we refer to walking on horizontal ground or in inclinations as level walking. The complete controller as in Fig. 4 is designed as a finite state machine with the four states:

- **Damp**: Moderate damping enabling “soft” foot down.
- **Free**: No or very small damping allowing free motion at stance.
- **Lock**: High damping which acts as a locking of the ankle at swing. But using a low damping force which can not entirely lock during stance.
- **Free Down**: No or very small damping allowing free motion during swing and stance.

All transactions between states, either for enabling or disabling damping, are based on the measured ankle angle and its motion $\Delta\alpha(t) = \alpha(t) - \alpha(t-1)$ where a positive $\Delta\alpha$ is a motion in dorsal direction. When switching between walking conditions i.e. stair descending and remaining the gait cycle maximum α_{max} and minimum α_{max} angle are used. Cycle start is defined as

$$Cycle\ start: \left\{ \begin{array}{ccc} \alpha(t) & = & \alpha^L \\ \Delta\alpha(t-1) & < & 0 \end{array} \right. \quad (1)$$

which corresponds to a foot lift. For each cycle are α_{max} and α_{min} observed.(Henceforth we will skip the sample index t for brevity.)

1) Stair Descending: In descending neither damping nor locking is preferable and only one state is necessary; **Free down**. During swing while moving the foot down to next stair case the toes point down for minimizing the time spent in swing. With an unlocked AFO angle a foot drop occurs caused by the shoe and foot weight. This helps in making a desirable motion.

Switching from level walking to descending occurs at *Cycle start* as (1) by detecting a downward motion and thereby avoiding ankle lock. A characteristic for descending is the large dorsi flexion needed. This is larger than at level gait even when using a large step. But due to the mechanical limitations can a threshold not be used for classification (Fig. 2) since the maximum value is the same in both stair descending and when using large steps in level walking. This motivates the use of a special state *Free down* only for stair descending and instead detecting the start and stop of the descention (Fig. 4). By initiating the descention with a short step on level ground a small plantarflexion is observed. If this is followed by a large dorsi flexion the direction of motion can be defined as

$$down: \left\{ \begin{array}{ccccc} \alpha^P & < & \alpha_{min} & < & \alpha^L \\ \alpha_{max} & = & \alpha^D & & \end{array} \right. \quad (2)$$

The comparison to α^L is to distinguish from ascending stairs where $\alpha_{min} = \alpha^L$ if the toe is put down first. *Down* (2) only detects the first step when descending. The remaining possible combinations of min and max represents cases of continuous gait and define *not down*:

$$\left.\begin{array}{lcl}\alpha_{min} & > & \alpha^P \\ \alpha_{max} & < & \alpha^D\end{array}\right\} \Rightarrow \text{Short steps or stair ascending} \quad (3)$$

or

$$\left.\begin{array}{lcl}\alpha_{min} & = & \alpha^P \\ \alpha_{max} & = & \alpha^D\end{array}\right\} \Rightarrow \text{Large steps or stair descending} \quad (4)$$

and

$$\left.\begin{array}{lcl}\alpha_{min} & = & \alpha^P \\ \alpha_{max} & < & \alpha^D\end{array}\right\} \Rightarrow \text{Inclinations.} \quad (5)$$

Stair descending can be initiated by either the orthotic foot or the users opposite sound foot. With the sound opposite foot the *down* sequence is accomplished with orthosis still being horizontal level. Sometimes due to pain or sense of instability AFO users prefer to start descending with the sound foot on level ground. The wearer has then to softly move the AFO foot down (in the same manner as wearing a non active AFOs) with his thighs to the first stair case. This corresponds to a short step and followed by a large dorsi flexion switch to *free down* occurs.

The end of descending can simply be defined as being back on *flat* ground:

$$flat : \alpha_{max} < \alpha^D \quad (6)$$

This ankle motion is accomplished by a short step on level ground. When the orthotic foot reaches horizontal level first this switch occurs directly. But, when the opposite sound foot starts level walking this causes an unwanted foot drop which leads to a short step and *flat* detection.

2) Level Walking: At level walking the controller switches between *Lock-Damp-Free* which corresponds to locked, damped and unlocked controller output (black arrow track in Fig 4). The states and transition rules are defined as:

- **Lock:** During swing the ankle is "locked" by the damper. The ankle locking force is chosen slightly smaller than the force caused by a heel strike. As a result, a heel strike results in a forced ankle motion. Thus a heel strike is detected if

  $$heel : \Delta\alpha < 0 \quad (7)$$

  This initiates a transition to *Damp*.
- **Damp:** *Damp* ends when the whole foot has ground contact. Therefore after maximum plantar flexion and

  $$toe : \Delta\alpha > 0 \quad (8)$$

  a transition to *Free* occurs. In this state the ankle can be rotated with only a small resisting torque in either direction.
- **Free:** To guarantee toe-clearance in the swing phase the ankle is locked when passing the threshold α^L. To avoid confusion with descending this is secured by classifying the step as being *not down* (3)-(5).

3) Stair Ascending: During stair walking the initial stair contact is made with the frontal part of the foot without the need of damping. This is handled with only two states *Lock* and *Free* which switch according to red arrow track in Fig.4:

- **Lock:** As in level walking the ankle is fixed during swing. But foot to ground is started by the *toe* (8) part is brought down first where the foot is dorsi flexed and thereby interrupting the locking. Transition from to *Free* occurs instantly.
- **Free:** In the same way as in level walking the *Lock* start is defined at *cycle start* by a *not down* (3)-(5) motion.

Switch between ascending and level walking is handled by detecting how the ground contact is made *toe*(8) or *hill*(7).

IV. Measurement

A. Subjects and Procedure

Tests were done walking in a treadmill at various inclinations, up and down stairs. The test group consisted of three healthy younger (20 $<$age$<$ 30) females whose length was approximately 1.65 meters. All participants used the same orthosis and shoes. Before each test a 5 min training wearing the AFO was conducted resulting e.g. in the capability of walking without the use of a railing. Walking in treadmill was done at 0.5 m/s and 1.0 m/s at -4, -1 degrees downhill, horizontal and 1, 6 uphill. The stair consisted of 21 staircases each 16 cm high and 30 cm deep and each subjects walked at a self selected speed. The results for each gait situation was averaged over the test group and several steps.

B. Equipment

The orthosis was made of composite with one steel joint on each side of the ankle. The mechanical ankle endpoints where $\alpha^P = -8$ and $\alpha^D = 26$ degrees. Reducing the noise influence at threshold comparisons (1)-(8) allowed 2 standard deviations. All electronics were embedded into a small box attached to the AFO. The digital controller was implemented in a low cost, off-the-shelf, 40 MHz PIC18F microprocessor. The angle sensor was sampled at 50 Hz with a 10-bit AD-converter. The signal was smoothed by a first order analog low-pass filter with cut-off frequency of 50 Hz. The MR-damper was current controlled using an 8 bit pulse-width-modulated signal. Furthermore, to analyze the system performance online, a Bluetooth unit enabled wireless logging of data to a PC.

V. Results

A. Level Walking

The algorithm detects the different walking states as shown in Fig.5 and coincides with the behavior of the desired controller. The plantarflexion limitation was more dominant than anticipated. At moderate speeds was dorsi flexion not influenced by the mechanical limitation. With increasing speed both maximum dorsi flexion and the maximum plantarflexion increased. It was noticed that the time of damping at foot down was automatically adjusted with gait speed. The actual locking angle differed from α^L by small variation for different steps caused by the combination of a fast foot lifting and a slow

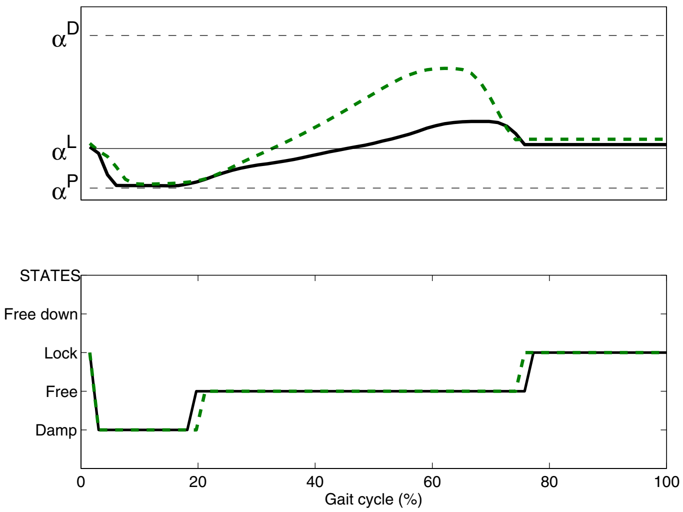

Fig. 5. Top; Averaged AFO angle using control at fast(green) and slow walking speeds(black). Bottom; Corresponding control states.

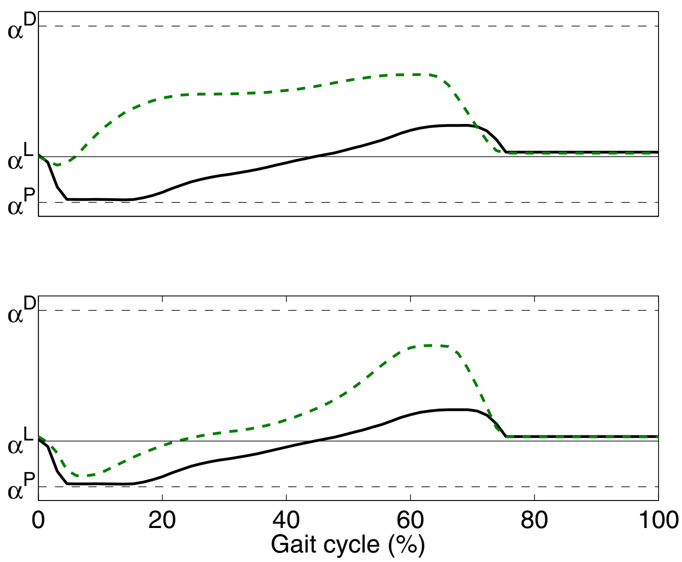

Fig. 6. Top: Averaged AFO angle using control, ascending a 6 degree inclination (green) compared with level walking (black). Bottom: Averaged AFO angle using control descending a 4 degree inclination (green) compared with level walking (black).

sampling. Since the test group had all sound feet, evaluation of the foot slap damping was not possible.

B. *Inclinations*

Walking in inclinations compensations were possible due to the low damping in the *Free* phase at stance. No significant differences where observed at small inclination angles $(-1, 1$ degrees).

A sound ankle adjusts both during stance and swing. With the controllable AFO, adjustments to changed inclinations are only possible during stance, see Fig. 6. This restriction makes toe damping period shorter for uphill gait heel strike compared to a sound foot. But adequate dorsi flexion is reached at stance.

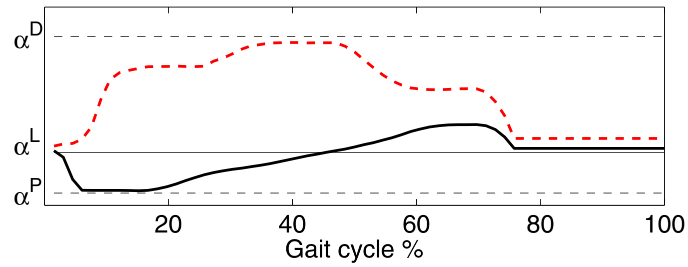

Fig. 7. Averaged AFO angle ascending stairs(red) compared to level walking (black).

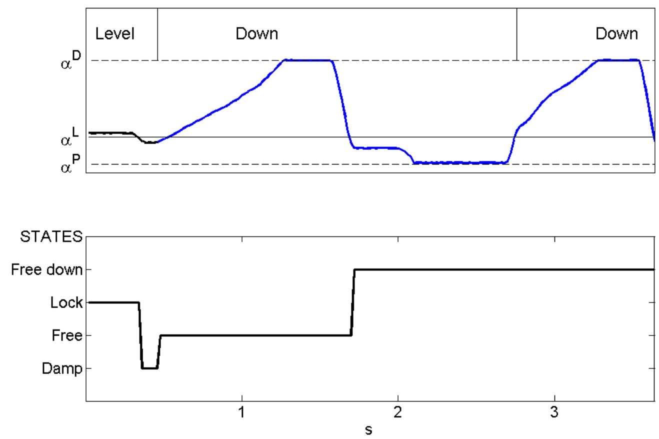

Fig. 8. Top: Resulting AFO angle using control when switching from level walking (black) to descending stairs(blue). Bottom: Corresponding control states.

At descending was the angle positively biased during the complete stance phase although a significant shorter step length not was observed. The increased dorsi flexion was also possible during the *Free* phase. It was also seen that the test persons sound ankle did not significantly influence the motion at foot lift as anticipated.

C. *Stairs*

The size of the mechanical endpoints is a choice of mobility or security. Here the α^D made undisturbed ascending at both foot down and foot lift possible (Fig 7). The limiting dorsi flexion was not reached for all in the test group. It is possible that the persons height compared to stair size influenced the size of flexion.

It was observed that during the lift phase the plantar flexion moment in some steps became to large for the locking mechanism. This made the algorithm switch to free stance to early and the toe-clearance was missed during this step. This occurrence depends of the weight and ankle strength of the user and choice of actuator. In descending the need for larger dorsi flexion capability increases (Fig.9). Although there was a mechanical small limitation the test group did not experience larger difficulties in walking. At foot lifting the sound muscles of the test group caused a delay of foot dropping. *Free down* enabled toe down motion to simplify the toe down in the

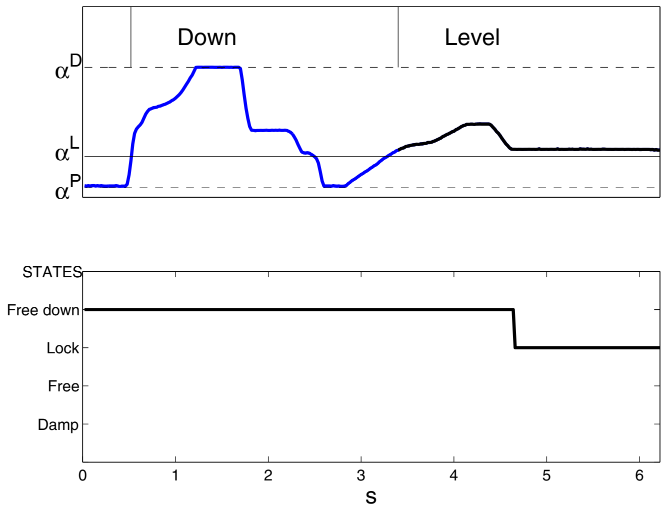

Fig. 9. Top: Resulting AFO angle using control when switching from descending stairs(blue) to level walking (black). Bottom: Control states.

initial stair contact. In the following forward progression of the lower limb was a new gait cycle initiated. The switching strategy between descending and level was shown to work at both stair start and ending. In the example shown in Fig. 8 descending is initiated with the sound opposite foot first. But the classification of down stair walking is not achieved until the end of the stance (appr. $t = 1.7s$). Meanwhile is the algorithm in the *Free* state. This does not cause problems to the wearer since the output the AFO ankle is the same *Free down* as in *Free* allowing a large dorsi flexion. In a similar manner is the ending of stair gait handled, see Fig. 9. Instead of switching to *Free* at foot down (appr. $t = 3.4s$) the algorithm continuous to stay in the *Free down* state until *Cycle start* is reached.

VI. Conclusions

A control system for active angle control of angle-foot-orthoses has been presented. With a magneto rheological damper the ankle damping during foot down and locking at swing was adjustable. Thereby was foot slap as well as foot drop avoided.

The controller was a 4-state machine with the states depending on gait phases and walking situations. Transitions from one state to another was based on the ankle angle. A single resistive sensor indirect measured this angle with a high signal to noise ratio. Using only three design parameters classification of level walking, ascending and descending stairs was possible. These parameters define the toe clearance angle during swing and maximum dorsi/plantarflexion. It was also shown that toe down in stair descending was made possible. The controlled AFO works independent of both gait speed and ground inclinations.

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