Adaptive Control of a Variable-Impedance Ankle-Foot Orthosis to Assist Drop-Foot Gait

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***Abstract*—An active ankle-foot orthoses (AAFO) is presented where the impedance of the orthotic joint is modulated throughout the walking cycle to treat drop-foot gait. During controlled plantar flexion, a biomimetic torsional spring control is applied where orthotic joint stiffness is actively adjusted to minimize forefoot collisions with the ground. Throughout late stance, joint impedance is minimized so as not to impede powered plantar flexion movements, and during the swing phase, a torsional spring-damper control lifts the foot to provide toe clearance. To assess the clinical effects of variable-impedance control, kinetic and kinematic gait data were collected on two drop-foot participants wearing the AAFO. For each participant, zero, constant, and variable impedance control strategies were evaluated and the results were compared to the mechanics of three age, weight, and height matched normals. We find that actively adjusting joint impedance reduces the occurrence of slap foot allows greater powered plantar flexion and provides for less kinematic difference during swing when compared to normals. These results indicate that a variable-impedance orthosis may have certain clinical benefits for the treatment of drop-foot gait compared to conventional ankle-foot orthoses having zero or constant stiffness joint behaviors.**

***Index Terms*—Actuator, drop foot, orthosis, rehabilitation.**

I. Introduction

Increasingly robotic technology is employed in the treatment of individuals suffering from physical disability, either for the advancement of therapy tools or permanent assistive devices. Initial research has focused primarily on devices that provide therapy to the arms of stroke patients [1]–[3]. However, lower extremity robotic devices have recently been developed [4]–[10]. When used for permanent assistance, adaptive orthoses enabled disabled persons to walk with greater ease and less kinematic difference when compared to normals [7], [8], [11]. Active leg prostheses also show promise. Preliminary studies report that the Otto Bock C-Leg, a microprocessor-controlled artificial knee, provides amputees with an increased independence compared with passive knee prostheses [12], [13].

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In this paper, a variable-impedance active ankle-foot orthosis (AAFO) is presented. The AAFO is designed to treat a gait pathology known as drop foot, a motor deficiency caused by total or partial central paralysis of the muscles innervated by the common peroneal nerve, or the anterior tibial muscle and the peroneal group. The two major complications of drop foot are slapping of the foot after heel strike (foot slap) and dragging of the toe during swing (toe drag). At heel strike, the foot generally falls uncontrolled to the ground, producing a distinctive slapping noise (foot slap) [14]. During midswing, toe drag prevents proper limb advancement and increases the risk of tripping [14].

A conventional approach to the treatment of drop-foot gait is a mechanical brace called an AFO. The use of AFOs has increased in popularity over the last several years [15]. Although AFOs offer some biomechanical benefits [16]–[20], disadvantages still remain. Carlson *et al.* [18] found that AFOs did not improve gait velocity or stride length in children with cerebral palsy. Still further, Lehmann *et al.* [19] discovered that although a constant stiffness AFO was able to provide safe toe clearance in drop-foot patients, the device did not reduce the occurrence of slap foot.

An active approach to the treatment of drop-foot gait is functional-electrical stimulation (FES). Here, short bursts of electrical pulses are applied to elicit muscle contractions [21], [22]. FES has shown some promise as a permanent assistance device [23]–[26], but the technology must be customized to the individual using trial-and-error methods and qualitative measurements. In most cases, a trained professional or clinician is necessary to qualitatively evaluate a subject's gait and incrementally change device settings.

Neither AFOs nor conventional FES systems adapt to the gait of the user, neither step-to-step changes in gait pattern due to speed or terrain, nor long-term gait changes due to changes in muscle function. In this paper, a computer controlled AAFO is presented where joint impedance is varied in response to walking phase and step-to-step gait variations. The AAFO comprises a force-controllable series elastic actuator (SEA) [27] capable of controlling orthotic joint stiffness and damping for plantar and dorsiflexion ankle rotations. We hypothesize that a variable-impedance orthosis has certain clinical benefits for the treatment of drop-foot gait compared to both unassisted gait and conventional AFOs comprising constant impedance joint behaviors [28]. Specifically, we anticipate that the major complications of drop-foot gait, namely foot slap and toe drag, can be reduced by actively controlling orthotic joint impedance in response to walking phase and step-to-step gait variations. Recent investigations have shown that for the healthy ankle-foot complex, ankle function during controlled

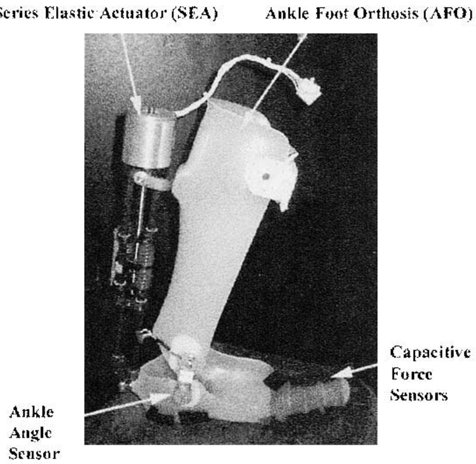

Fig. 1. AAFO is shown. An actuator, attached posteriorly to a conventional AFO, varies orthotic joint impedance based on position and force sensory information.

plantar flexion closely resembles a linear torsional spring where ankle moment is proportional to ankle position [29]. Thus, we anticipate that by adjusting the stiffness of a virtual linear torsional spring acting about the orthotic joint, forefoot collisions can be minimized and the slap foot complication alleviated, not only at a single speed, but at every forward walking speed. Furthermore, during swing, we expect that a proportional derivative (PD) control applied to the orthotic joint, with gains that vary with gait speed, will dorsiflex the ankle through a greater angular range to provide sufficient clearance at all walking speeds. Finally, for individuals suffering from unilateral drop-foot gait, we hypothesize that changing orthotic joint impedance will result in a more symmetric gait between affected and unaffected legs.

In this paper, pilot kinetic and kinematic gait data are collected on two drop-foot participants walking at slow, self-selected, and fast speeds. For each participant, zero, constant, and variable impedance control strategies are evaluated, and the results are compared to the mechanics of three age-, weight-, and height-matched normals.

II. Methods

A. AAFO

To investigate different AFO control schemes and their effect on drop-foot gait, an actuator and sensors were attached to a conventional AFO. The device, shown in Fig. 1, has a total weight of 2.6 kg (excluding the weight of an off board power supply). In the sections to follow, the actuator, AFO, sensors, and control system are described.

1) SEA: The SEA, previously developed for legged robots [27], [30], [31], was used to control the impedance of the orthotic ankle joint for sagittal plane rotations. The SEA consists of a brushless dc motor in series with a spring. The SEA provides force control by controlling the extent to which the series spring is compressed. The deflection of the spring was measured by a linear potentiometer sampled at 1000 Hz and passed through a first order filter with a cutoff frequency equal to 50 Hz. The signal was then numerically differentiated and passed through another first order filter with a cutoff frequency of 8 Hz. The deflection of the series spring was controlled using a PD controller.

The advantages of the SEA are that it has low impedance, the motor is isolated from shock loads, and the effects of backlash, torque ripple, and friction are filtered by the spring [27]. A further advantage is that the SEA exhibits stable behavior while in contact with most environments, even when in parallel with a human limb. For this study, the SEA allowed for the implementation of any virtual, torsion mechanical element about the ankle.

2) AFO: A standard polypropylene AFO with a metallic hinge (Scotty) ankle joint was specifically fabricated to fit study participants [32]. This joint allowed free motion in the sagittal plane (plantar and dorsiflexion), but was rigid for inversion/eversion movements. The AFO was modified by molding two recesses one at the heel and the other at mid-calf. In these recesses, several holes were drilled to attach the SEA (Fig. 1).

3) Ankle Angle Sensor: A Bourns 6637S-1-502 5-kΩ rotary potentiometer was used to determine the angle between the shank and the foot. The angle sensory signal, sampled at 1000 Hz, was passed through a first-order low-pass filter with a cutoff frequency of 50 Hz. The ankle velocity was found by differentiating the pot signal and then passing it through a second-order Butterworth filter with a cutoff frequency of 8 Hz [33].

4) Ground Reaction Force (GRF) Sensors: To measure GRFs, an Ultraflex system [34] was used. Here, six capacitive force transducers, 25 mm square and 3 mm thick, were placed on the bottom of the AFO, two sensors beneath the heel and four beneath the forefoot region. Each sensor could detect up to 1000 N, had a resolution of 2.5, and a scanning frequency of 125 Hz. The signal from each sensor was passed through a first-order filter with a cutoff frequency equal to 5 Hz.

A single foot switch, model MA-153 [35], was placed in the heel of a shoe worn with the orthosis to detect heel strike approximately 30 ms earlier than the Ultraflex force sensors.

5) Electronics: Using 12 m of cabling, the AAFO was connected to a CIO-DAS08/JR-AO analog and digital I/O card in a PC running Debian Linux and a Kepco JQE 36-15 power supply for the SEA. The signals were processed by code written in C and then transmitted to the amplifier of the SEA.

B. Adaptive Controller for Drop-Foot Gait

1) States and Triggers: A finite-state machine was implemented to address each complication of drop-foot gait. Three states were used each with a specific control objective (Fig. 2). In a Contact 1 state, from heel strike to midstance, the objective of the controller was to prevent foot slap. During a Contact 2 state, from midstance to toe-off, the controller minimized the impedance of the brace so as not to impede power plantar flexion movements. Finally, in a Swing state, spanning the entire swing phase, the user's foot was lifted to prevent toe drag. A Safe state was created to shut off the device when any unexpected circumstances occurred. The triggers or transitional parameters for the finite state machine are shown in Fig. 3.

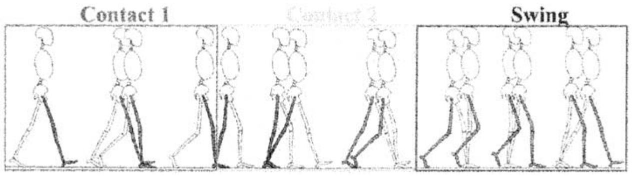

Fig. 2. Individual states for the finite-state machine. Contact 1 spanned the first half of ground contact from heel strike to the middle of midstance when the tibia first became perpendicular with the foot. Contact 2 spanned the second half of ground contact, beginning when the tibia first became perpendicular with the foot and ending at toe-off when the leg first lost contact with the ground. Finally, the Swing state spanned the entire swing phase, from toeoff to heel strike.

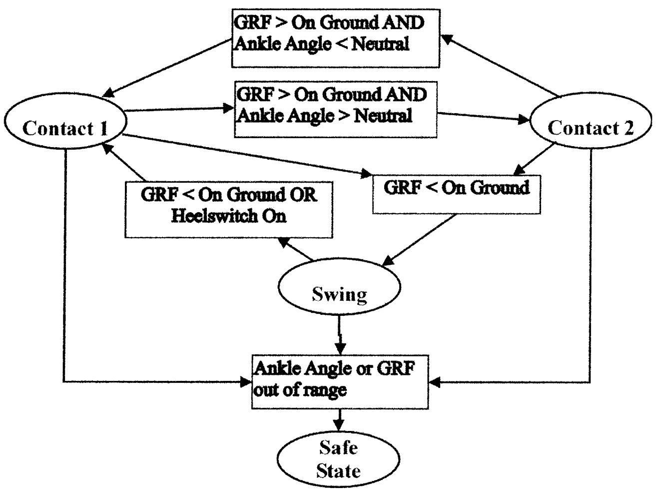

Fig. 3. Triggers for the finite-state machine. For a typical gait cycle, Contact 1 began when the foot switch within the heel was compressed. The transition into Contact 2 occurred when the GRF, equal to the sum of all six force transducers, was greater than On Ground, equal to 60 N, and when the ankle was in dorsiflexion. The ankle was considered to be in dorsiflexion when the angle between the tibia and foot was less than 90°. Furthermore, On Ground was set to 60 N because this particular value reliably discerned ground contact from noise during swing. Contact 2 ended when the GRF was less than On Ground. In fact, the transition into Swing always occurred when the GRF was less than On Ground. The controller transitioned to the Safe State when any of the force or angle sensory signals went beyond a specified normal operating range. The acceptable range for each force sensor was 1000 N, the maximum force that any one sensor should measure in walking for a 90-kg person. The acceptable range for the angle sensor was ±45°, the normal operating range for the human ankle.

2) *Control Algorithm:*

a) Contact 1: A recent investigation [29] has shown that during controlled plantar flexion (CP), normal ankle function can be modeled as a linear rotational spring where ankle moment is proportional to ankle position. Thus, during the CP phase of walking, a linear torsional spring control was used for the orthotic ankle joint. As a criterion for selecting the preferred stiffness of the orthotic torsional spring, the controller analyzed the ground reaction force generated at the moment of forefoot impact after each walking step. For this investigation, the extent of foot slap was deemed too extreme, and the CP stiffness too low, if a high-frequency force spike occurred at the moment of forefoot collision.

In Fig. 4, a representative forefoot force signal from a drop-foot participant is compared to a forefoot force signal from a normal participant. Both participants wore the AAFO under a zero impedance control, and the forefoot force signal was computed from the sum of all four force transducer signals

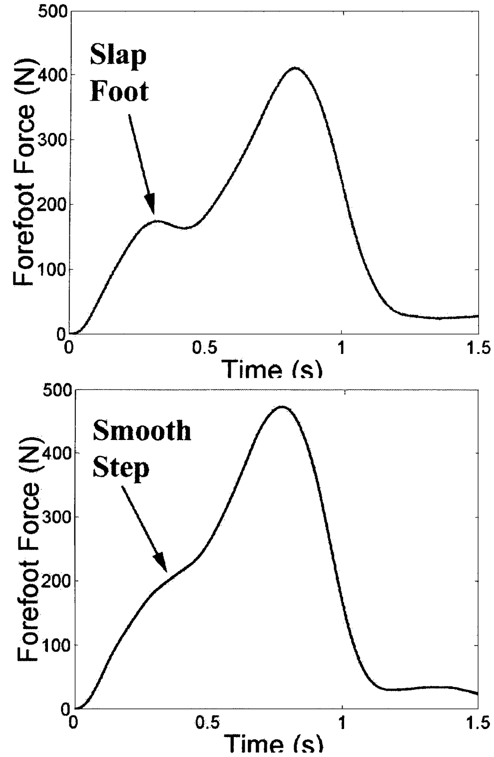

Fig. 4. Comparison of the forefoot ground reaction force from (a) a drop-foot participant and (b) a normal participant. In (a), a high-frequency force spike is shown (arrow) indicating the foot slap complication of drop-foot gait. In (b), no force spike exists indicating a smooth heel to forefoot transition in the normal participant.

TABLE I
RULES FOR CHANGING THE ANKLE STIFFNESS DURING CP

Number of slaps in last 5 steps (n)	Change in Ankle Stiffness
0	$-\Delta\Gamma$
1	0
2-5	$(n-1)\,\Delta\Gamma$

measured in the forefoot region (Fig. 1). In Fig. 4(a), a dual peak force pattern indicated the occurrence of foot slap in the drop-foot participant, whereas in Fig. 4(b), the lack of a dual force spike was an indication that no foot slap had occurred in the normal participant.

To detect the dual peaks and the occurrence of foot slap, the AAFO controller numerically differentiated the forefoot force and then filtered that signal using a second-order Butterworth filter with a cutoff frequency of 0.6 Hz. If substantial foot slap occurred, the differential of the forefoot force was found to be negative, and the stiffness of the orthotic torsional spring stiffness was incremented. The CP stiffness started at zero and was incremented by the rules shown in Table I, where the incremental stiffness $(\Delta\Gamma)$ was 5.7 Nm/rad (0.1 Nm/deg), approximately 2% of the anticipated final ankle stiffness [36].

Gait speed is an important step-to-step gait variation for which the AAFO should respond and adapt. In this paper, the time of foot contact, defined as the time that a foot remains in contact with the ground from heel strike to toe-off, was used as a measure of forward speed. With an expectation that orthotic

CP stiffness should change with gait speed, the full range of gait contact times were divided into bins, denoting velocity ranges. During each swing phase, stance time was estimated from the orthotic force transducers, and the participant's time of contact bin, or forward speed range, was selected. Within each bin, the AAFO controller optimized the orthotic CP stiffness. It was determined that only three bins were necessary to span the full speed range of the participants. In a previous study of above-knee prostheses control, the time of contact was also used to estimate forward gait speed [37].

b) Contact 2: The drop-foot participants did not experience any difficulties during powered plantar flexion. Hence, the control objective of Contact 2 was to minimize orthotic joint impedance so as not to impede the participants' power plantar flexion movements. During this state, the SEA's target force was set to zero.

c) Swing: During the swing phase, a second-order under-damped mechanical model (spring-damper PD control), previously used to characterize normal ankle function [29], [36], [38], [39], was used to control the orthotic ankle joint. Using the AAFO, each drop-foot participant walked at slow, self-selected, and fast speeds, and the swing-phase ankle angle was collected on both the affected and unaffected sides. At each speed, orthotic joint stiffness was increased manually until the early swing phase dorsiflexion velocity measured on the affected side matched the unaffected side. Orthotic joint damping was increased from zero until unwanted joint oscillations were removed. The final values of stiffness and damping are listed in Table II. The stiffness and damping values for the drop-foot users were not correlated with gait speed directly, but with ranges of stance time, in the same manner to the CP stiffness control described earlier.

TABLE II
ANKLE STIFFNESS AND DAMPING VALUES USED IN THE AAFO DURING SWING TO PREVENT TOE DRAG

Gait Speed	K (Nm/rad)	B (Nms/rad)
Slow	28.65	0.57
Normal	37.24	1.03
Fast	45.84	1.15

TABLE III
SUBJECTS' SEX, AGE, ANTHROPOMETRIC DATA, AND GAIT SPEED

Subject	Sex	Age (yr)	Mass (kg)	Height (m)	Self-Selected Gait Speed (m/s)
Drop Foot	M	62	79.1	1.79	1.22
Drop Foot	M	62	95.4	1.77	1.07
Normal	M	66	76.6	1.70	1.39
Normal	M	67	86.1	1.75	1.01
Normal	M	67	73.2	1.70	1.22

C. Clinical Evaluation of the AAFO

1) Subjects: The clinical evaluation of the AAFO was conducted in the Gait Laboratory, Spaulding Rehabilitation Hospital, Boston, MA. Protocol approval was provided by the Spaulding Rehabilitation Hospital and Boston University School of Medicine institutional review boards. Moreover, a written informed consent was obtained from each participant before data collection began.

Drop-foot participants having only a unilateral drop-foot condition were selected, and on their affected side, participants did not suffer from a gait disability other than drop foot. Both participants had an absence of strongly manifesting spasms and contractures in lower extremity joints. Finally, each participant had used an AFO for at least two years and, therefore, was experienced at AFO ambulation. Subjects reached a stable neurological state after the incident that caused their disability. Thus, no recovery of function was expected or found. Three normal subjects were matched for gender, height, weight, and age to the drop-foot participants. Subject sex, age, mass, height, and self-selected gait speed are listed in Table III.

2) Data Collection: Kinematic and kinetic data were measured on both the affected and unaffected sides using an eight-camera VICON 512 system [40] and two AMTI force plates [41]. The data were processed at 120 Hz with VICON Workstation [42] using the standard model of the lower limbs included with the software [43]–[45]. These data were then analyzed using MATLAB [46].

3) Test Procedure: The subjects donned the AAFO in three different control conditions: zero, constant, and variable impedance. The zero impedance control scheme was implemented by setting the target force on the SEA to zero, thereby minimizing the impedance contribution of the orthosis across the ankle joint. This scheme was meant to approximate unassisted drop-foot gait. For the constant impedance control scheme, the AAFO controller commanded a constant joint stiffness, independent of walking phase and gait speed. This joint stiffness was the converged CP flexion stiffness from the variable impedance control that minimized the number of slap foot occurrences at the self-selected gait speed. This constant impedance control condition was designed to imitate conventional AFO technology employed in the treatment of drop-foot gait [28].

For each controller, subjects walked at slow, self-selected and fast gait speeds. The subjects first walked at their self-selected speed using the constant impedance control scheme. The amount of time required to cover a specified distance was measured using a stopwatch. Subjects were then asked to reduce their time by 25% for the fast gait speed and increase their time by 25% for the slow gait speed. These times were then matched when testing the remaining two control conditions.

4) Method of Analyzing Data:

a) Normalization to stride cycle: A stride cycle was defined as the period of time for two steps, and was measured from the initial heel contact of one foot to the next initial heel contact of the same foot. All data were time normalized to 100% of the stride cycle. The ankle angle data during a gait cycle were fitted with a cubic spline function and then resampled to 200 samples so that each point was 0.5% of the gait cycle.

b) Gait symmetry: In this study, it was assumed that normal gait was symmetrical and that deviations from a reference pattern were a sign of disability. To analyze spatial asymmetry, the step length on the affected side (L_{affected})

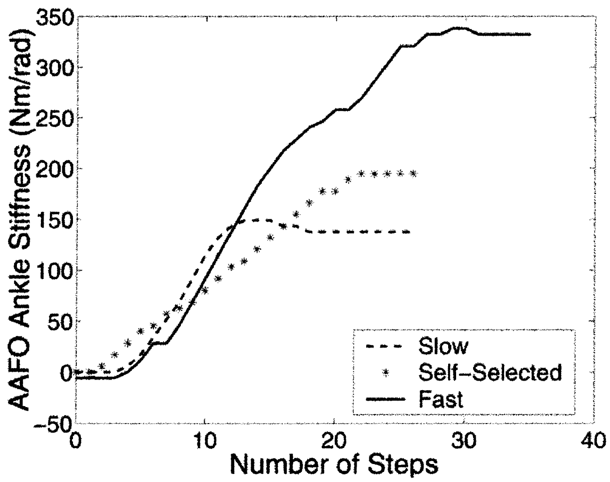

Fig. 5. Orthotic joint stiffness is plotted against the number of steps taken by a participant starting from an initial default impedance value of zero. At each gait speed, the AAFO controller converged on a final stiffness value after only 32 steps. As shown, the CP stiffness increases with increasing gait speed.

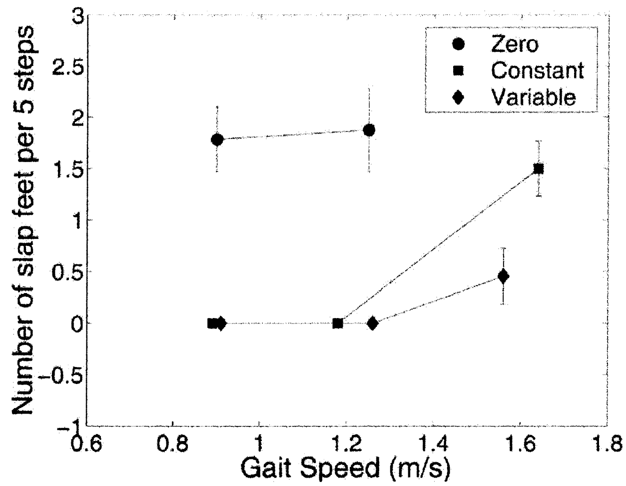

Fig. 6. Slap foot occurrences per 5 steps ($n = 5$) measured on two drop-foot subjects walking at slow, self-selected, and fast speeds. The three curves correspond to the zero, constant and variable impedance control conditions.

was subtracted from the step length on the unaffected side ($L_{\text{unaffected}}$). The difference in stride lengths (L_{sym}) should be zero for symmetric gait

$$L_{\text{sym}} = L_{\text{affected}} - L_{\text{unaffected}}. \tag{1}$$

To analyze temporal asymmetry, the step time on the affected side (T_{affected}) was subtracted from the step time on the unaffected side ($T_{\text{unaffected}}$). The difference in stride times (T_{sym}) should be zero for symmetric gait

$$T_{\text{sym}} = T_{\text{affected}} - T_{\text{unaffected}}. \tag{2}$$

c) Statistical methods: A multiple comparison using a one-way ANOVA was used to determine which means were significantly different for the gait symmetry. P values less than 0.05 were considered significant for all tests.

III. Results

The first evaluation of the drop-foot controller was to test whether the system was capable of converging to a final CP stiffness that reduced or prevented slap foot, as described in the Methods Section II-B-II-A. For each of the three gait speeds, the controller was able to converge to a final stiffness value within 32 steps (Fig. 5). During the stiffness convergence at each of the three gait speeds, the occurrences of the high frequency forefoot force signal [typical of slap foot; see Fig. 4(a)] were reduced.

As a measure of the slap foot complication, the average number of occurrences of slap foot per five steps (25 steps total) were calculated for each drop-foot subject, control condition, and gait speed ($n = 5$). The participants were unable to walk at the fast gait speed using the zero force condition because it was not deemed safe. The constant impedance condition eliminated the occurrences of slap foot at the slow and self-selected gait speeds (Fig. 6). However, slap foot occurrences increased at the fast gait speed. By adjusting CP stiffness with gait speed in the variable-impedance control condition, the number of occurrences of slap foot was reduced at the fast gait speed by 67% compared to the constant stiffness condition.

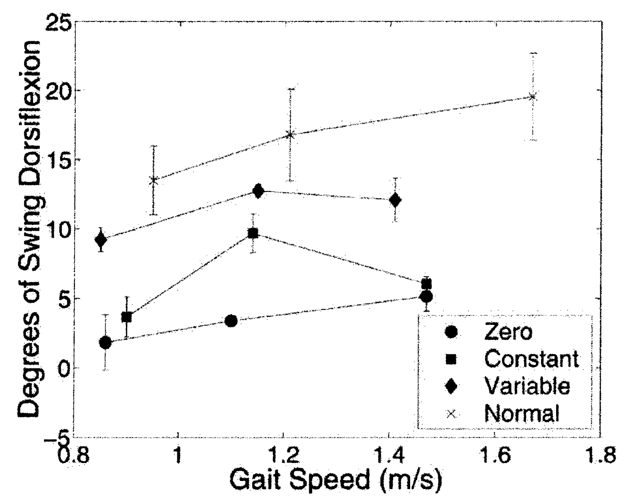

Fig. 7. Amount of swing dorsiflexion for normal ($n = 3$) and drop foot ($n = 2$) participants are plotted. All data points for the normal participants are an average of 15 trials, whereas for the drop-foot participants the averages are over 20 trials.

To quantify the reduction of the second major complication of drop foot, or toe drag, the swing dorsiflexion angular range was used. The dorsiflexion angular range was defined as the maximum plantar flexion angle during the powered plantar flexion phase of stance minus the maximum dorsiflexion angle during swing. The variable impedance control was able to increase the amount of swing dorsiflexion as compared to the constant impedance condition by 200%, 37%, and 108% for slow, self-selected, and fast gait speeds, respectively (Fig. 7).

A constant impedance AFO should hinder powered plantar (PP) flexion since a dorsiflexion moment will be exerted against the foot during late stance. As expected, the constant impedance condition reduced the PP angle as compared to the zero impedance condition and the normals (Fig. 8). Here, the PP angle was defined as the maximum plantar flexion angle during power plantar flexion minus the maximum dorsiflexion angle during controlled dorsiflexion in stance. The variable-impedance controller had a larger PP angle than the

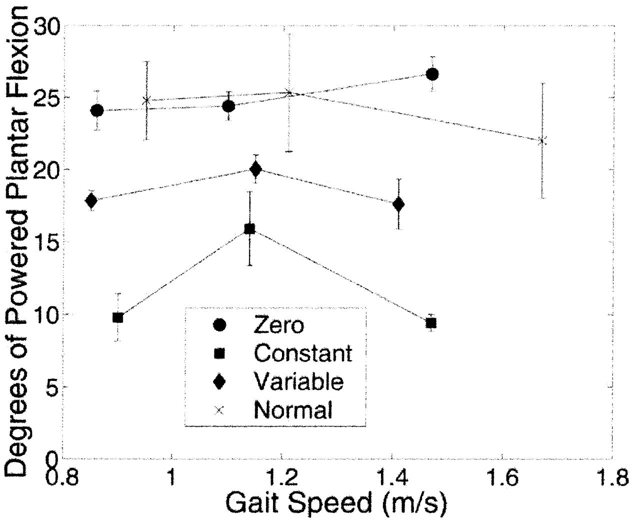

Fig. 8. Amount of powered plantar flexion for normal ($n = 3$) and drop foot ($n = 2$) participants are shown. All data points for the normal participants are an average of 15 trials, whereas for the drop-foot participants the average is over 20 trials.

TABLE IV
DIFFERENCE IN STEP LENGTHS $(L_{\mathrm{SYM}})(M)$ AND STEP TIMES $(T_{\mathrm{SYM}})(S)$ BETWEEN THE AFFECTED AND UNAFFECTED SIDES

	L_{sym} (m)		T_{sym} (s)	
	Self-selected	Slow	Self-selected	Slow
Zero Impedance	0.08 ± 0.07	0.09 ± 0.09	0.09 ± 0.07	0.15 ± 0.16
Constant Impedance	0.04 ± 0.06	0.02 ± 0.08	0.07 ± 0.05	0.04 ± 0.12
Variable Impedance	0.02 ± 0.07	0.00 ± 0.07	0.02 ± 0.09	0.01 ± 0.16

constant impedance control condition by 89%, 25%, and 82% for the slow, self-selected, and fast gait speeds, respectively.

To evaluate spatial and temporal gait symmetry, the differences in step lengths $(L_{\mathrm{sym}})(m)$ and step times $(T_{\mathrm{sym}})(s)$ from the affected to the unaffected side were compared for each of the three control conditions ($n = 20$) (Table IV). Both L_{sym} and T_{sym} for the variable-impedance controller were significantly smaller than the zero impedance controller for both the self-selected and slow gait speeds ($p < 0.05$). The zero and constant impedance conditions were significantly different for the slow gait speed ($p < 0.05$). For the fast gait speed, a comparison was not possible because the step length for both sides could not be calculated for a single walking cycle.

IV. Discussion

Drop-foot gait is caused by stroke, cerebral palsy, multiple sclerosis and neurological trauma from accident or surgical complication [47]. Although drop foot is a common pathology, with over 250 000 cases in the U.S. from stroke alone [48], current ankle foot orthoses are nonadaptive and fail to eliminate significant gait complications [18], [19]. In this paper, we present an active AFO, and we evaluate zero, constant, and variable-impedance control strategies on two persons suffering from unilateral drop-foot gait. We find that actively adjusting joint impedance in response to walking phase and forward speed reduces the occurrence of slap foot, and provides for swing phase ankle kinematics more closely resembling normals as compared to the zero and constant impedance control schemes. Furthermore, we find a variable-impedance control allows for greater PP flexion compared to a conventional constant stiffness approach where a dorsiflexion spring impedes PP flexion movements during late stance.

A. Gait Symmetry

Although the major complications of drop foot are reduced with a variable-impedance control, our findings do not support the hypothesis that changing orthotic joint impedance will result in a more symmetric gait between affected and unaffected legs in unilateral drop-foot gait. To test the hypothesis, we evaluated spatial and temporal gait symmetry according to the difference in step lengths and times between affected and unaffected sides. When using the variable-impedance control, the difference in step time and step length was not significantly different from that measured with the constant impedance control condition. However, for both gait speeds analyzed, the variable-impedance controller did improve spatial and temporal gait symmetry compared to the zero impedance control condition, whereas the constant impedance control did not.

B. Intercycle Impedance Variations: Adapting CP Stiffness to Gait Speed

The CP stiffness was optimized within each gait speed range or time of contact bin. After the variable-impedance controller adapted CP stiffness across gait speed, the final stiffness at the slow speed was 36% less, and at the fast speed, 57% greater than at the self-selected speed. Thus, from slow to fast speeds, stiffness increased more than twofold. A constant stiffness spring tuned only to the self-selected speed allowed slap foot to occur at fast walking speeds (Fig. 6). It also made the ankle too stiff during slow walking, reducing the angular rotation of the ankle during CP movements in early stance.

C. Intercycle Impedance Variations: Adapting Swing Impedance to Gait Speed

The primary concern for both the drop-foot participants in the study was catching their toe during swing and loosing their balance. With constant swing phase impedance, both users caught their toe at the fast gait speed. This was not surprising given the fact that, for normal gait, the amount of time to lift the foot and achieve toe clearance was found to decrease by a factor of two from slow to fast speeds. To achieve this time decrease with the AAFO, a fourfold increase in swing joint stiffness was necessary (Table II). Thus, changing orthotic joint impedance with gait speed, in order to lift the toe during swing, appears to be an essential control feature of the variable-impedance AAFO.

D. Intracycle Impedance Variations

Normal ankle function has been modeled as a linear spring during CP, and as a nonlinear, stiffening spring during controlled dorsiflexion [29]. Throughout the swing phase, the ankle has been represented by a linear torsional spring and damper [49]. Given these differences in ankle function within a single gait cycle, an assistive ankle device, acting in parallel with the human ankle-foot complex, should ideally change its impedance in response to walking phase. To this end, a state controller was used in the AAFO, and joint impedance was modulated in response to walking phase. During the CP flexion phase of walking, or Contact 1, a linear torsional spring control was employed where the stiffness was adjusted to prevent slap foot. From midstance to preswing, or the Contact 2 state, a zero impedance control was implemented so as not to impede normal PP flexion movements. Finally, during the Swing state, a spring-damper PD control was implemented to provide toe clearance. In this study, the primary difficulty with the constant impedance control was the reduction of PP flexion movements (Fig. 8). Here, the spring-damper control used to prevent toe drag was acting against the foot when the users attempted to plantar flex their ankle during late stance.

E. PP and Force Sensor Resolution

The variable-impedance controller should have a similar maximum PP angle as the zero impedance condition since both controllers were designed to *not* impede late stance PP movements. However, in this study, this behavior was not observed (Fig. 8). It was discovered that the variable-impedance controller transitioned into the Swing state too early, before the foot actually left the ground, due to a lack of resolution in the forefoot force sensors. Consequently, the Swing spring-damper controller was activated too early, impeding PP movements during late stance. To solve this problem in future investigations, we feel a foot switch might have to be positioned in the forefoot region to more accurately detect the event of toe-off.

F. Drop-Foot Participants' Feedback

In the development of any rehabilitation device, including AFOs and FES systems, feedback from the user is always an important consideration. Despite the weight of the AAFO and its current bulkiness, both drop-foot participants preferred the device over any AFO they had ever used. They expressed a desire to use the AAFO at home and wanted to be informed of any commercialization plans. One participant remarked that the AAFO made walking "almost subconscious, like normal walking."

G. Future Work

Before the variable-impedance AAFO can have broad utility for individuals suffering from drop-foot gait, system hardware and software must be advanced. The series-elastic actuator used in this investigation is too heavy and power intensive to be practical in a commercially available active ankle-foot system, and would, therefore, have to be redesigned. Still further, control strategies and sensing architectures specifically suited for the ascent and descent of stairs and ramps would be necessary. In addition to these improvements, different variable-impedance AAFO controllers might be developed for therapy purposes. In this study, the drop-foot controller allowed one slap foot for every five steps taken by the user, and reduced the stiffness when no occurrence of slap foot occurred. For someone requiring permanent assistance, like the drop-foot participants in the study, the final stiffness provided sufficient support to where the participants could not discern foot slap. However, the drop-foot controller could also be used as a therapy tool to promote improvements in muscle function and control. In this case, the controller might be adjusted to allow more than one slap foot per five steps, thus providing the minimal support for gait and promoting the development of muscle dorsiflexors.

V. Conclusion

In this paper, we demonstrate that a variable-impedance control applied to an AAFO reduces the dominant complications of drop-foot gait in two drop-foot participants. By actively adjusting joint impedance, the occurrence of slap foot was reduced and the swing phase ankle kinematics more closely resembled normals as compared to the zero and constant impedance control schemes. In the development of AFO to treat drop-foot gait, we feel modulating orthotic joint impedance in response to walking phase and gait variation is an important design goal.

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