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An Ankle Based Soft Active Orthotic Device Powered by Pneumatic Artificial Muscle

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Abstract— **Soft robotics are made by materials which have similar modulus with human musculoskeletal system. They can be used to augment the human performance without restricting the natural behavior. This paper presents a bio-inspired, ankle-based soft active orthotic device which can assist the ankle dorsiflexion during walking. This device implemented a silicone-based, fast actuating Pneumatic Artificial Muscle (PAM) to provide angular assistant force at the ankle joint. This PAM is based on the pneumatic network structure. Specific design have been made to make the PAM ergonomically compile with foot-ankle structure and facilitate the underlining application. An initial testing was first carried out to characterize the PAM. The control strategy was planned based on ankle angle information within each gait cycle. A pilot study was carried out for evaluation. The results show that this soft active orthotic device can improve the dynamic stability of the ankle joint. This device can be potentially used as real time argumentation for frail and fall-prone elderly and benefit their walking stability.**

I. Introduction

Wearable robotics have been implemented in many areas such as the human robot collaboration, assistive augmentation, and rehabilitation engineering [1]. Traditionally, wearable robotics are made of rigid materials such as steel and aluminum. One typical application is the exoskeleton, which has been mainly used in lower limb rehabilitation and argumentation [2]. Functionally, the lower limb exoskeletons can assist locomotion for those who lost their walking ability or aid in rehabilitation after stroke or serious surgery [3]. However, exoskeleton systems are bulky, heavy, and in most cases, expensive. More importantly, they are not comfortable to be wore and can result in injuries of human body due to misalignment or control errors. Thus, lower limb exoskeletons were seldom used to assist frail people, who might partially lose their ability to keep balance due to aging process or balance-compromising diseases such as Parkinson's disease and diabetes.

In recent years, soft robotics have become a fast-emerging research topic. Especially, the wearable soft robotic systems have been proposed to augment or assist the natural performance of human-being. They were mainly applied in the military and medical rehabilitation area [4-7].

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Wearable soft robotics are made by materials which have similar modulus with human musculoskeletal system. Thus, they can augment the human performance without restricting natural kinematics. The actuation systems of those are often pneumatic or hydraulic. They are safer to interact with human body. Besides, the structures of the wearable robotics can be customized with technology such as 3D print. Therefore, it can fit to human body segment. And the cost can be much lower than the conventional exoskeletons.

Many wearable soft robotic systems for lower limbs have been proposed as laboratory prototypes [4-7]. For example, Ding et al. introduced a hip-based mono-articular soft exosuit coupled to a lab-based multi-joint actuation platform that can aid the hip extension [4]. Jin et al. presented a soft robotic suit that can provide a small yet effective assistant force for hip flexion through an elastic winding belt [5]. Wan et al. introduced a fluidic actuator design that can generate compliant and powerful actuation at the knee joint. This device can provide knee augmentation at a low operating pressure [6].

Ankle joint plays a vital part in the balance control during standing and walking. Park et al. [7] introduced a wearable robotic device powered by pneumatic artificial muscle (PAM) actuators for ankle–foot rehabilitation. The PAM design in their study was like the McKibben actuator, which provides linear force when the PAM was shortened due to the pressure change. They designed the wearable system to help the ankle dorsiflexion and plantarflexion for the dropped-foot patients during walking.

In this paper, we present a bio-inspired, ankle-based soft active orthotic device which can assist the ankle dorsiflexion during walking. This device implemented a silicone-based, fast actuating PAM to provide angular assistant force at the ankle joint. This PAM is based on the pneumatic network structure (i.e., pneu-net) proposed by Mosadegh et al. [8]. Specific changes in the PAM design have been made to make them ergonomically compile with foot-ankle structure and facilitate the underlining application. We first carried out to an initial testing to characterize of this PAM. Then, we designed the control strategy based on ankle angle information within each gait cycle. Lastly, a pilot study was carried out to evaluate whether this active orthotic device can improve the dynamic stability of the ankle joint angle.

II. The System Design

A. System Overview

Fig. 1 shows the system overview of the ankle-based soft active orthotic device. The device has an integrated pneumatic system and control system. The pneumatic system has an air compressor (800W-30L, Outstanding Co., China) as the air source. It can supply the compressed air ranged from 0kPa to

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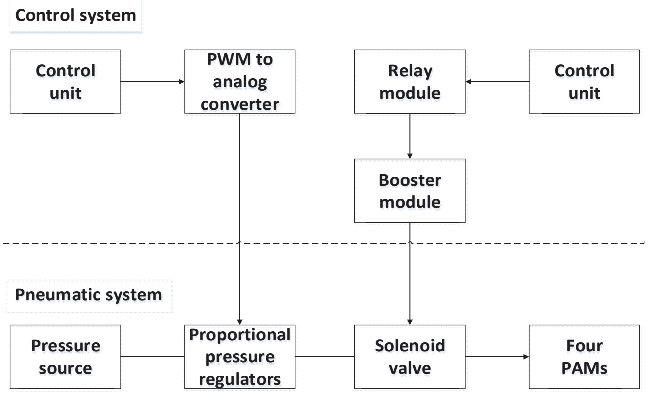

Figure 1. The system overview

800kPa. It has a manual valve to control the air flow and an analog gauge to display the output pressure. To maintain desired and constant pressure, the air compressor is connected to a proportional pressure regulator (VPPE-3-1-1/8-10-420-E1, Festo, Germany) which can be manipulated by the signals from a control unit. It is then connected to a 3-way/2-position normally closed solenoid valve (M-Type-DC24, High-end Pneumatic Co., China) which controls the inflation or deflation of four PAMs (two for each foot).

The control system has two individual control units. Both control units are made based on an Arduino-Uno R3 system-on-chip. The first control unit generates a Pulse Width Modulation (PWM) signal which is later converted to a voltage signal by a PWM-to-analog converter (0-10V, STIME Co., China). This adjustable analog signal is used to control the proportional pressure regulator. Another control unit receives the information of the ankle angle within each gait cycle, and based on that, generates corresponding control signals. It is connected to a relay module (LB16, Yunhui Co., China) to send signal to switch on/off the solenoid valve to determine the inflation and deflation of the PAM. Meanwhile, a booster module (XL6009-DC-DC, Yunkai Co., China) is also used in the circuit to increase the power supply (5V) to 24V to make the solenoid valve function properly.

B. The Pneumatic Artificial Muscle (PAM)

The most important part of this active orthotic device is the PAM which deforms according to pressure change and provide angular assistant force at the ankle joint. The PAM is a silicone-based soft actuator powered by a pneumatic system. It is designed based on the fast pneu-net structure proposed by Mosadegh et al. [8]. It mainly consists of two layers, i.e. an extensible layer (top layer) and an inextensible but flexible layer (bottom layer). Both layers were made by silicone.

The top layer is extensible because it consists of multiple air chambers connected by one single channel. The chambers are closed to each other with small gaps in-between. Each chamber has its individual walls. Specific design has been made to let the walls between each chamber thinner than the top and bottom walls. This design allows the larger expansion along horizontal axis of the PAM than that along the vertical axis. Thus, during the pressure-driven deformation, there is only small change in the height of the PAM but large expansion.

Although the bottom layer is also made by silicone, it is reinforced by embedding a piece of paper. This changes the material property, which makes the bottom layer relatively inextensible.

When the pressurized air is supplied through the channel into the air chambers of the top layer, those air chambers expand and stretch against each other. At the same time, the bottom layer does not extend as much as the top layer do. As a result, the PAM will bend against the bottom layer due to the different compliance properties of the two layers. This bending mechanism generates exerted force, which allows the PAM to provide assistant force at the ankle joint.

In order to make this PAM suitable for our specific application, there were a few design considerations. First, in order to provide real time assistance during every gait cycle (each gait cycle takes 800ms to 1200ms), the PAM need to deform rapidly. As indicated by Mosadegh et al. [8], the deformation rate depends on the volume of the air chambers. In other words, fewer air chambers could result in more rapid deformation. However, more air chambers are also preferred in a way that it can result in larger exerted force, which might be more helpful to keep the walking stability. Second, for the sake of wearability, the PAM need to be small, light weighted, yet provide enough force when deformation. Thus, the overall height, width, thickness of the PAM, and the design of air chambers need to be balanced. Bear that in mind, specific changes were made compared to the initial design of Mosadegh et al. The design specification of the presenting PAM was summarized in Table 1.

TABLE I. DESIGN SPECIFICATION OF THE PAM

Specification	Measurement
Overall Height, Length, Width of Top Layer	21 mm, 225mm, 20 mm
Height, Length, Widths of AC	17 mm, 8mm, 20 mm
Inner Height, Length, Widths of AC	15 mm, 4mm, 16 mm
Top/ In-between Wall Thickness	3mm, 2mm
No. of Air Chambers	20
Height, Length, Width of Bottom Layer	4 mm, 225mm, 20 mm

AC- Air Chamber

The fabrication procedure of the PAM was simple. The top and bottom layers were made by pouring the liquid elastomer (Ecoflex 30, Smooth-On Inc., PA, US) into the molds (Fig. 2)

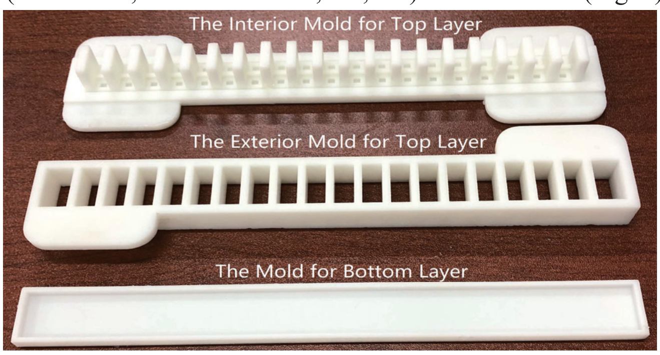

Fig 2. The 3D printed molds

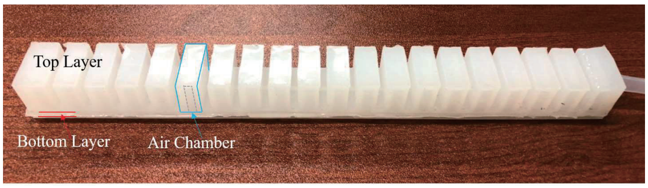

Fig 3. The fabricated PAM

and allowing them to cure (i.e. become the rubber form) under room temperature. The top and bottom layers were fabricated separately and then glued together.

The top layer was made with an interior and an exterior mold. The interior mold has the structure to allow the elastomer to form the air chambers, whereas the exterior mold consists of parallel plates that results in the formation of the walls between each air chamber. The bottom layer was fabricated by a single mold. Before the liquid elastomer was poured in, a piece of paper which fit the size the mold was inserted, to reinforce the bottom layer.

Those molds were first designed in Solidworks (Solidworks Co., MA, US) based on the above-mentioned design specification, and later fabricated by a 3D printer (Lite 300HD, UnionTech, CN) based on the Stereo lithography Appearance (SLA) technology. Fig. 2 shows the 3D printed molds.

After pouring the liquid elastomer, the molds were placed in a vacuum chamber to draw the bubbles (resulted from the pouring) to the surface. And a needle was used to pop those bubbles. The liquid elastomer would be cured after 4 hours under the room temperature. Then the PAM was removed from the molds and the top and bottom layers were glued together. One end of the top layer was pricked with a tiny hole and an air tube was used to connect the PAM to the air source. Fig. 3 shows the PAM fabricated.

The PAM was inserted into a closely tightened textile wrapping made by flexible fabric (Fig. 4). The wrappings with PAM were then sewed on the lateral and medial sides of the ankle joint based on a customized foot orthotic brace. This design can facilitate the exerted force generated by the PAM to be transmitted by the orthotic brace and provide additional toque to the joint ankle.

C. The Initial Testing

The purpose of the initial testing was to characterize the minimum pressure required for the PAM to deform completely, and the deformation rate (i.e. how fast the PAM can achieve complete deformation). Six different pressure levels were tested, including 33kPa, 44kPa, 55kPa, 76kPa, 87kPa and 100kPa.

Fig 4. The active orthotic device with two PAM for each foot

Fig 5. The PAM deformation under different air pressure

The PAM deformation under four selected air pressure was depicted in Fig. 5. The PAM was considered as completely deformed if the distal end touched itself. The results showed that the PAM can achieve complete deformation if the supplied air pressure was above 55kPa. As the pressure increased, the deformation rate will increase. In other words, the PAM can deform more rapidly, which is preferred. However, once the applied air pressure reached to 87kPA, further increasing of the air pressure might not change the deformation rate much further. Since the PAM is expected to have longer service life (i.e. more inflation-deflation cycles) under lower air pressure, 87kPa was chosen as the working pressure for this PAM. The mean time for complete deformation under 87kPa for 100 inflation-deflation cycles was 267±24ms. With such rapid deformation, the PAM can deform and recover according to the change of gait parameters within one gait cycle.

D. Control Strategy

The actuation of the active orthotic relies on the angular deformation of the PAM, which in turn assist the ankle dorsiflexion when it bends. The control strategy of the PAM was based on the ankle plantarflexion and dorsiflexion angle within each gait cycle. Thirteen participants were involved in a walking experiment to obtain such information.

During the experiment, a metronome was used provide a predefined tempo. Each time, the participants were asked to follow the metronome's tempo in a way that each (left and right) heel strike coincided with the beat. Two different beating frequency were tested, i.e. 48 Bpm(beat/min.) and 60 Bpm. Thus, for each walking trial, the participants had a fixed cadence. And each gait cycle lasted for 2.5 second (48 Bpm) and 2 second (60 Bpm), respectively. Each participant walked three times under one metronome frequency.

An eight-camera motion capture system (VICON, Oxford Metrics, Oxford, UK) was used for gait analysis. Thirty-two reflective markers were placed strategically on the bony landmarks according to VICON's Full Body Plug-in-Gait Model [9]. The plantarflexion and dorsiflexion angle of the

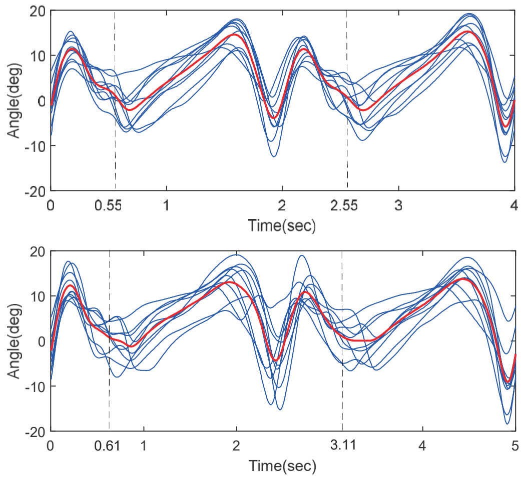

Figure 6. Ankle joint profile under different walking cadence, the dash line indicates one gait cycle

ankle joints were calculated based on the Plug-in-Gait Model. The timing of heel contact was determined by the minimum vertical trajectories of the heel markers. The ankle joint profile within one gait cycle was determined by the mean angle profiles of the 13 participants. Fig. 6 shows the ankle joint profile of the left foot under different walking speed.

The PAM inflation and deflation were determined by the ankle angle joint profile. In order to assist the ankle dorsiflexion, the PAM need to inflate and deform when the ankle angle decreases. Whereas, it needs to deflate and recover when the ankle angle increase, so that it won't hinder the plantarflexion. Based on this, the control strategy of the PAM was determined and demonstrated in Fig. 7.

III. Experiment

A pilot study was carried out to evaluate the effect of the active orthotics on the walking stability. One participant was involved. The participant was asked to walk on an instrumented treadmill (Fig. 8). During walking, he was asked to try to place each heel strike in accordance to the tone of the metronome. Similar to the previous experiment, two levels of walking cadence were considered, i.e. 1/2 s^{-1} and 1/2.5 s^{-1}. For

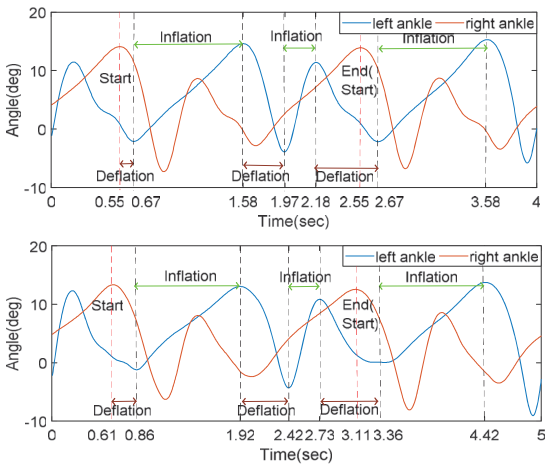

Figure 7. The control strategy of the PAM

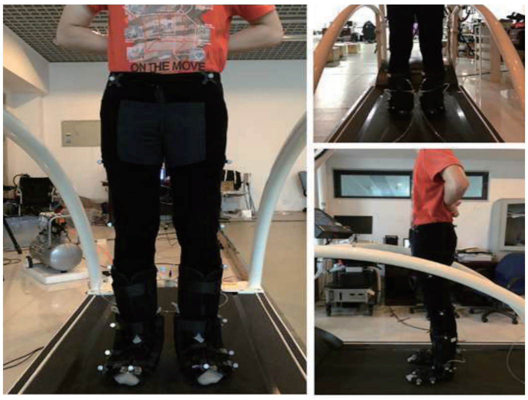

Figure 8. Marker placement and experiment set-up

each walking cadence, the speed of the treadmill was 1.1 km/h and 1.5 km/h respectively. An eight-camera motion capture system (VICON, Oxford Metrics, Oxford, UK) was used to capture the walking kinematics. This study was approved by the university's Medical Ethical Review Board.

There were three testing conditions, namely, no orthotics (NO), wear orthotics without actuation (WN), wear orthotics with actuation (WA). During the WA condition, the active orthotics was controlled based on the above-mentioned control strategy for each walking cadence level.

The maximum finite time Lyapunov exponent (λ_{max}) for the ankle joint angle (left/right) was used to quantify the local dynamic stability during walking. λ_{max} represents the average logarithmic rate of divergence of a system that generates close trajectories [10]. The divergence is used to quantify the stability of a dynamical system in response to small perturbations. Thus, it has been widely used to quantify the stability of the "human walking system" [10, 11]. As smaller λ_{max} indicates a faster divergence, which means a better joint angle stability, the hypothesis of the presenting study is that under the WA testing condition, the walking will be more stable, result in a smaller λ_{max}.

IV. Results

The maximum finite time Lyapunov exponent (λ_{max}) were calculated from 50 consecutive gait cycle (starting from the 20th. gait cycles after the walking initiated). An example of the divergence curve was shown in Fig. 9. The divergence when wearing the orthotic device is faster, indicating higher dynamic stability. Table II shows the λ_{max} under different testing conditions. The results demonstrated that wearing the active orthotics can result in a reduced λ_{max}, indicating an improvement on the local dynamic stability. The actuation of the PAM further reduced the λ_{max} at the left ankle (1/2 s^{-1}), indicating it might further improve the walking stability.

TABLE II. The Maximum Finite Time Lyapunov Exponent

Cadence /Direction	NO	WN	WA
1/2 s^{-1} R	1.52	1.47	1.49
1/2 s^{-1} L	1.56	1.59	1.53
1/2.5 s^{-1} R	1.25	1.21	1.21
1/2.5 s^{-1} L	1.29	1.21	1.27

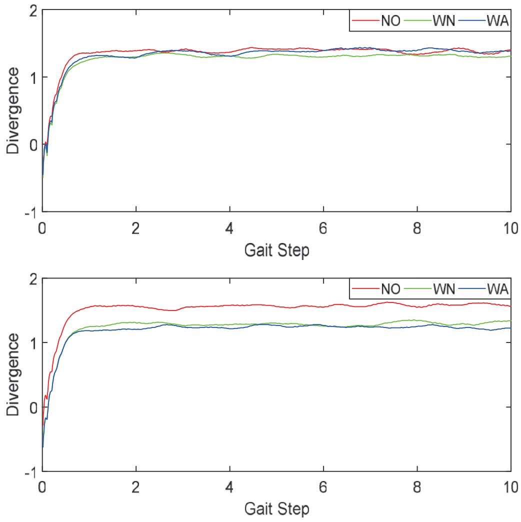

Figure 9. The divergence curve under different testing condition

V. Discussion

This paper presented a novel design of an ankle based soft active orthotic device which can assist the ankle dorsiflexion during walking. The preliminary results from the pilot study demonstrated that it can improve the local dynamic stability of the ankle joint during walking. Thus, this device can be potentially used as the real time wearable argumentation for frail and fall-prone elderly and benefit their walking stability.

Aging effect or balance-compromising diseases can degrade an individual's ability to maintain a stable walking gait. Previous research has demonstrated that instability during walking can be an indicative measurement of higher risk of falling [12]. Even though elderly often adopt to a slower walking speed to increase their walking stability, it is often found that slow walking strategy is associated with higher gait variability [13]. Previously, elderly people were encouraged to conduct the strength and balance training to increase their walking stability. However, this type of proactive training might not be always effective. Afterall, aging is an irreversible process, and the degeneration of muscle strength and sensory system is inevitable. The existing heavy and bulky lower limb exoskeleton systems are for those who completely lost their locomotion ability or have pathological gait patterns. To our knowledge, there was lack of device that can reactively strengthen the walking stability for aging and frail people.

The active orthotic device was actuated by a pair of PAMs, which are strategically placed at the ankle joint. The PAM is light-weighted and soft. Therefore, it is more compatible to be used to argument the motion of human body segment. The PAM generated angular force/toque at the ankle joint. Unlike the wearable soft robotic system introduced by Park et al. [7], where the PAM provided linear force, the design in the presenting study can provide more intuitive and direct assistance at the ankle joint.

To accurately evaluate gait stability, it is necessary to consider not only how a single stride was performed, but also how movements are controlled from one stride to another. This requires continuous monitoring of gait. Therefore, in this study, the local dynamic stability was quantified by maximum finite time Lyapunov exponent from a non-linear dynamics approach. The examination of this parameter revealed that the ankle joint stability is improved by the active orthotics, However, future research needs to carried out to assess the detailed change of gait kinematics and kinetics.

The control strategy of this device was based on the gait information obtained from experiment study. Future work needs to be carried out to implement wearable sensing technology to realize the real time feedback control of this device.

This device presented in this study aimed to help the frail and fall-prone elderly with their walking stability. However, as a pilot study, young healthy participant was involved for the proof of concept. Future research needs to be carried out to evaluate the effect of device on elderly with a larger sample size.

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