A Soft Robotic Intervention for Gait Enhancement in Older Adults

Xinyao Hu, Xu Zeng, Yang Xu, Chuang Luo, Liyao Jia, Zhong Zhao, Zhenglong Sun, *Senior Member, IEEE*, and Xingda Qu

***Abstract*—Falls continue to be a major safety and health concern for older adults. Researchers reported that increased gait variability was associated with increased fall risks. In the present study, we proposed a novel wearable soft robotic intervention and examined its effects on improving gait variability in older adults. The robotic system used customized pneumatic artificial muscles (PAMs) to provide assistive torque for ankle dorsiflexion during walking. Twelve older adults with low fall risks and twelve with medium-high fall risks participated in an experiment. The participants were asked to walk on a treadmill under no soft robotic intervention, inactive soft robotic intervention, and active soft robotic intervention, and their gait variability during treadmill walking was measured. The results showed that the proposed soft robotic intervention could reduce step length variability for elderly people with medium-high fall risks. These findings provide supporting evidence that the proposed soft robotic intervention could potentially serve as an effective solution to fall prevention for older adults.**

***Index Terms*—Falls in older adults, fall risks, gait enhancement, intervention, soft robotics.**

I. Introduction

FALLS continue to be a major safety and health concern for older adults. In the US, falls are the leading cause of both fatal and non-fatal injuries among people aged 65 and above [1]. Due to the aging population worldwide, there has been an increasing trend in fall injury incidences in recent years [2]. Besides physical injuries, falls also have negative psychological consequences such as fear of falls, anxiety, and depression [3], [4]. These psychological consequences can make older adults lose independence in their daily living.

Impaired gait has been recognized as an important risk factor for falls among older adults [5], [6]. Some researchers have attempted to reduce fall risks by developing interventions aimed at improving gait performance [7]–[9]. Exercise and training interventions have been most widely used in this regard [10]. Wang *et al.* [11], for example, examined the effects of a 12-week exercise intervention comprised of resistance, endurance, and balance training. Their results showed that the exercise intervention positively affected gait endurance and gait performance in both normal-speed and fast-speed walking conditions among older adults. More recently, van der Straaten *et al.* [12] have introduced a visually augmented gait training which involves treadmill walking plus visual feedback of gait information. They found that this training intervention could be effective in reducing tripping-related falls, as it led to increased minimum toe clearance during gait in both younger and older adults. A limitation with the exercise and training interventions is that they require professional physical therapists to provide guidance and assistance, which are labour intensive, time-consuming and expensive.

With recent advances in robotic technology, lower-limb robotics have been developed to assist human walking [13]. Lower-limb exoskeletons that can provide weight-bearing functions are often used in gait rehabilitation, especially for those who partially or completely lose their walking ability [14]. However, lower-limb exoskeletons are typically bulky and heavy, and can even result in body injuries due to misalignment or control errors [15]. These limitations restrict the wide acceptance of lower-limb exoskeletons by older adults with impaired gait and high fall risks [16].

Over the past decade, wearable soft robotics have become a fast-emerging research topic. Soft robots are typically made of the materials that exhibit intrinsic compliance and elasticity (e.g., textile and polymeric elastomers) [17]. Thus, they can be applied without restricting natural kinematics during body movement. More importantly, soft robots are safer to interact with than the traditional rigid-bodied exoskeletons [18]. Therefore, soft robots could be a more practical and useful intervention for gait enhancement. Many wearable soft robotic prototypes have been proposed and applied in gait enhancement. The most notable examples are the series of soft exosuits designed by Harvard Biodesign Lab [19]–[22].

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This work involved human subjects or animals in its research. Approval of all ethical and experimental procedures and protocols was granted by the Institutional Review Board of 334 Shenzhen University under Approval No. 20190012.

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These exosuits are made of low-profile textiles materials, and have been used in various gait assistance applications, such as loaded gait assistance [21] and gait rehabilitation for stroke patients [20]. Other examples of applications of soft robots in gait enhancement include Jin *et al.* [23] who developed a soft robotic suit to facilitate hip flexion, and Sridar *et al.* [24] who proposed a soft-inflatable exosuit to assist swing-phase knee extension for post-stroke patients' gait rehabilitation.

Recently, some soft robots specifically designed for ankle assistance have been reported. For example, Chung *et al.* [25] developed a robotic boot with soft and inflatable pneumatic artificial muscles (PAM) to provide assistive torque for ankle plantarflexion during walking. Kim *et al.* [26] developed an ankle-foot orthosis (AFO) based on a pneumatic actuating system. The AFO consists of a portable and highly effective air compressor which can generate pressurized air up to 800 kPa to provide dorsiflexion assistance to hemiplegic patients with drop-foot. Thalman *et al.* [27] also developed a soft AFO, i.e., Exosuit, for drop-foot assistance. This Exosuit had two types of actuators. The thermally bonded nylon actuators were used to assist ankle dorsiflexion, and variable stiffness soft actuators were used to enhance ankle joint proprioception during stance phase. Similar design of variable stiffness soft actuators has been used for the correction of foot inversion and eversion of stroke patients during aquatic rehabilitation as well [28].

Ankle joint plays a vital role in balance control during gait. Ankle dorsiflexion can affect both foot placement at heel contact and ground clearance at toe off, which are critical kinematic factors for gait balance [29]. Kemoun *et al.* [30] found that elderly people with a history of falls had delayed dorsiflexion response both kinematically (joint angle) and kinetically (joint moment) during the swing phase. Age-related muscle weakness of tiabilis anterior (i.e., the main dorsiflexor) was found to be associated with poor foot clearance during swing phase, which might increase the risk of tripping and falls [31]. Thus, the proposed soft robotic intervention was designed to assist dorsiflexion during gait.

Gait variability reflects the fluctuation of gait parameters (e.g., step length, stance time) from one step to the next [32]. Increased gait variability among older adults is often caused by impaired dynamic balance control during gait [33]. Researchers have reported that increased gait variability is associated with increased fall risks [5], [6]. For example, Maki [5] reported that step width variability can be prospectively predictive of all-cause falls.

Thus, the objective of the present study was to develop and evaluate a novel soft robotic intervention for ankle dorsiflexion enhancement among older adults. Different from previous studies. an ankle-based soft actuators were designed with customized PAMs. The effects of the soft robotic intervention on gait variability were examined. Both low-fall-risk and medium-high-fall-risk older adults were included. We hypothesized that the proposed soft robotic intervention could help decrease gait variability in both the low-fall-risk and medium-high-fall-risk older adults, but its effects would be different between groups. We hypothesized that the proposed soft robotic intervention could help decrease gait variability in both the low-fall-risk and medium-high-fall-risk older adults, but its effects would be different between groups.

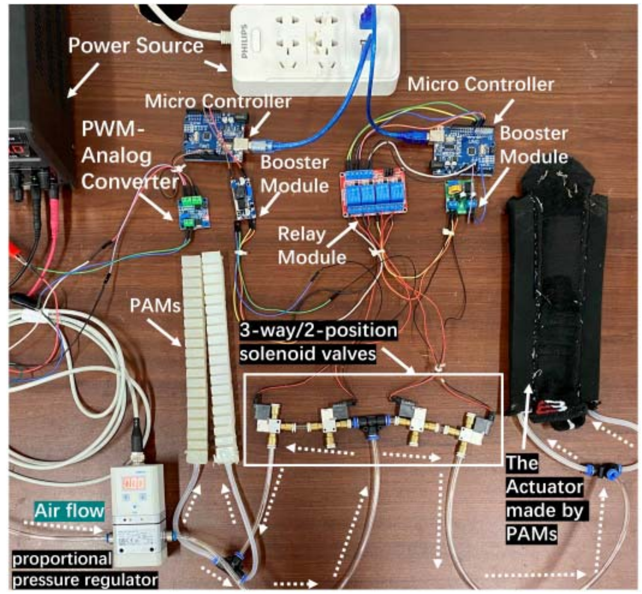

Fig. 1. The soft robotic intervention system.

II. Materials and Method

A. Soft Robotic Intervention System

1) System Overview: The soft robotic intervention system (Figure 1) is actuated pneumatically. An air compressor (800W-30L, Outstanding Co., China) is used to supply compressed air with air flow rate at 60 L/min. The compressed air goes through a proportional pressure regulator (VPPE-3-1-1/8-10-420-E1, Festo, Germany) whose function is to maintain constant air pressure level at 110 kPa. The pressure regulator is connected to the 3-way/2-position solenoid valves (M-Type-DC24, High End Pneumatic Co., China), and the solenoid valves are connected to the customized PAMs. The pneumatic parts are all connected with plastic air tubes with 3mm inner diameter. The control hardware consists of two micro controllers that are custom-made based on Arduino-Uno R3. The first micro controller sets the output pressure level for the proportional pressure regulator through a PWM-to-Analog converter (STIME Co., China) based on the pre-programmed pulse width modulation (PWM) signal. Another micro controller is connected to a relay module (LB16, Yunhui Co., China) and sends signals to control the 3-way/2-position normally closed solenoid valves, which in turn controls the timing of PAM inflation and deflation. The inflation and deflation mechanism is shown in Figure 2. Two solenoid valves are connected to direct the air into the PAMs at each side of the leg. When the solenoid valve A is energized and solenoid valve B is de-energized, valves A2, A3, B1, B3 are open, and the PAM is inflating. When both the solenoid valve A and solenoid valve B are de-energized, valves A1, A3, B1, B3 are open, and the PAM maintains the inflation status. When the solenoid valve A is de-energized and solenoid valve B is energized, valves A1, A3, B2, B3 are open, and the PAM is deflating. All the components in the soft robotic intervention system, except the air compressor (220V) and the proportional

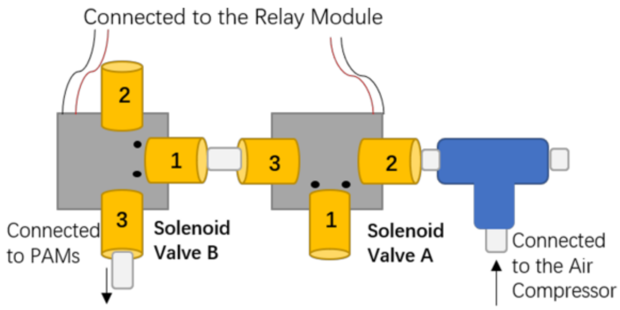

Fig. 2. The inflation and deflation mechanism.

pressure regulator (24V), are powered through its USB series port (5V). Two booster modules (XL6009-DC-DC, Yunkai Co., China) are used to amplify the power supply from 5Vto 24V to make the solenoid valves function properly.

2) Design of the Actuator: The PAMs, which are considered as the actuator of the system, are customized based on the following design considerations. First, the PAM should generate sufficient torque to assist ankle joint dorsiflexion. This soft robotic intervention was aimed for gait enhancement for those who might be frail but still capable of walking. Moreover, there was lack of extant literature regarding how large the assistive torque should be to improve their gait. Thus, the initial design goal was tentatively set to allow the actuator to generate maximum torque no less than 3 Nm, which was approximately 50% of the maximum isokinetic dorsiflexion torque of the ankle for elderly people with body weight of around 60kg [34]. The maximum torque of the actuators in the present study was estimated to be 3.64 $\pm$ 0.27 Nm, which should be sufficient to assist ankle joint dorsiflexion. Second, the actuation should be fast enough to allow the desired inflation/deflation to be completed within the time frame of one complete gait cycle. The timing of inflation and deflation are both important considerations. The former determines the actuation speed (i.e., how fast the assistive torque can be provided to the ankle), and the latter determines how quickly the assistive torque can be dismissed so that it does not hinder the subsequent plantarflexion. Compared to the motor or cable driven system, the speed of actuation can be an inherent limitation of pneumatically powered actuators. Thus, the current intervention system was designed to provide assistance for slow walking speed only, with targeted cadence levels from 48 steps/min to 72 steps/min. These selected cadence levels were equivalent to approximately 60% to 90% of the mean cadence for elderly who had low physical activity intensity level [35]. Such design could allow desired actuator inflation/deflation to be achieved within the time frame of one complete gait cycle. Third, the PAM should be light-weight and small-size to its greatest extent so that it cannot hinder natural gait kinematics.

To address these considerations, the PAM used here is based on the fast pneu-net structure proposed by Mosadegh *et al.* [36]. In particular, the PAM mainly consists of two silicone layers, i.e., an extensible layer (top layer) and an inextensible but flexible layer (bottom layer). As shown in Figure 3, the bottom layer is reinforced by embedding a piece of paper, which makes the bottom layer relatively inextensible.

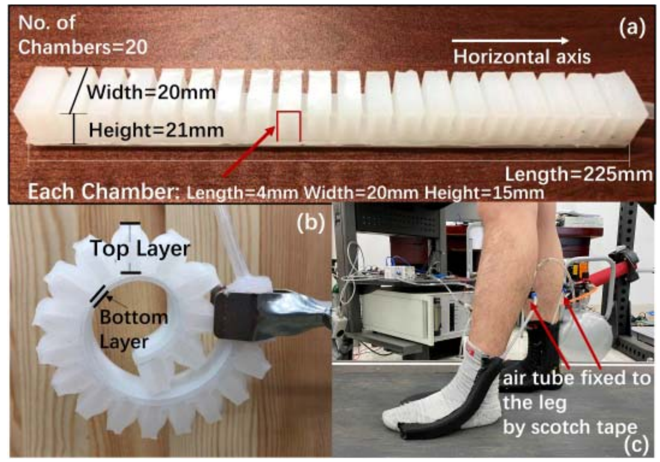

Fig. 3. (a) The design of the PAM; (b) Inflated PAM; (c) Actuators worn around the ankle.

The top layer consists of multiple air chambers connected by a single channel. When the pressurized air is supplied through the channel into the air chambers, the air chambers expand and stretch against each other. The walls between chambers are made thinner than the top and bottom walls. This allows the range of chamber expansion to be greater along the horizontal axis than along the vertical axis. The bottom layer cannot extend as much as the top layer. As a result, the PAM can bend against the bottom layer and generate controlling torques. The bending speed and resultant torque are dependent on the design of walls, the size, and number of the air chambers. The detailed characterization of this PAM can be found in Mosadegh *et al.* [36], and the final design and specification are shown in Figure 3(a). The manufacturing of the PAM follows the procedure introduced in Mosadegh *et al.* [36]. The PAM was first inserted into a pocket which was made of elastic textile and its shape and size were customized according to the PAM. Two PAM pockets were sewn on the medial and lateral sides of a sports sock, and the distal end of the PAMs was approximately aligned with the metatarsal joint. During the intervention, the PAM-attached socks were worn by the participants as shown in Fig. 3(c). The total weight of the actuator (i.e., the part that attached to the ankle) was 76.6g.

Initial tests showed that the PAM could be fully actuated (when the curvature shape of the PAM was no longer changing with the increased pressure level) when the operation pressure was set at 87kPa or above [37]. However, too high pressure may make the PAM break. The pressure level was determined at 110 kPa by trial-and-error tests. Specifically, the PAMs were tested under different pressure levels separately, and the one under the level of 110 kPa did not break at the end of the four-hour test (approximately 15000 cycles). Thus, the operation pressure level was set at 110kPa.

The torque generated by the actuator was assessed with a blocked displacement characterization test (Figure 4(a)). One end of the actuator was affixed on a rack. The other end of the actuator inflated against a digital force gauge (range = 50N, resolution = 0.01N, sampling rate = 10Hz). The air flow was controlled to allow continuing inflation for 600 ms, and then deflation for 400ms. This procedure was repeated for 100 cycles. The left and right PAM was tested separately.

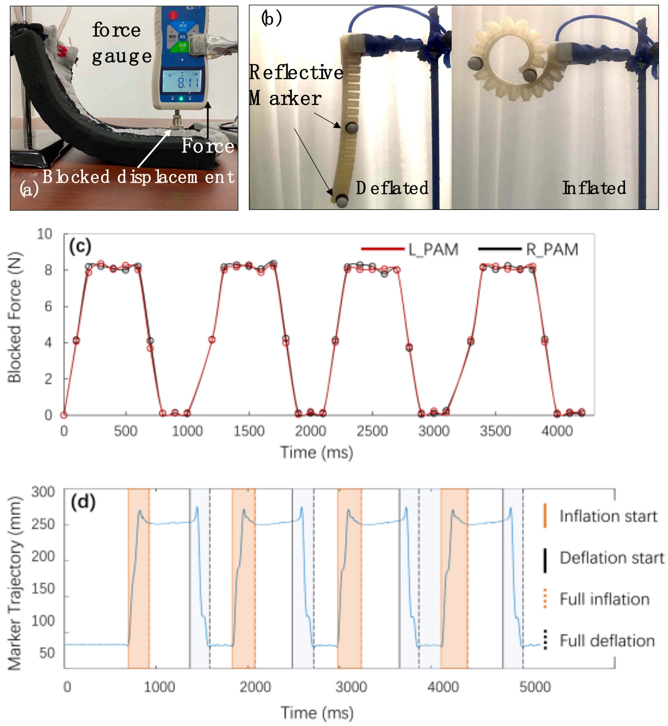

Fig. 4. (a) The blocked displacement characterization test of the PAM; (b) The test for the inflation/deflation speed; (c) The resulted blocked force of the PAM (showing four inflation/deflation cycles), L_PAM = PAM on the left side, R_PAM = PAM on the right side; (d) The vertical trajectory of marker at the far end of the PAM (during the four inflation/deflation cycles).

The average blocked force of each PAM was 8.09 ± 0.61 N (mean ± std). Given that the length of the PAM is 0.225m, the estimated maximum torque of the actuators (generated by the two PAMs at each ankle) was 3.64 ± 0.27 Nm.

To determine how fast the PAM could fully inflate and deflate under the pressure level of 110kP, the PAM was vertically placed with one end anchored firmly on a rack and the other end free to move in the air, where the PAM deformed against its own weight (Figure 4(b)). Two reflective markers were placed on the side of the actuator. The weight of each reflective marker was approximately 3g. Thus, the effect of their weights could be neglected. An eight-camera motion capture system (VICON, Oxford Metrics, Oxford, UK) was used to track PAM motion at a sampling rate of 200 Hz. The vertical trajectory of the far-end marker was used to measure the timing of PAM deformation. Since the length of the tube which connected the solenoid valve and PAM was only 55 cm, it was assumed that the PAM started to inflate immediately after the solenoid valve was open. The testing results showed that the full inflation time under the operation pressure was 222 ± 24 ms, and full deflation time was 202 ± 14 ms.

3) Control Strategy: A pre-programmed control strategy based on the reference ankle joint angle profile was developed and saved in the controller to control the timing of PAM inflation and deflation. The timing of PAM inflation and deflation was determined based on the ankle joint angle profile instead of the joint torque profile because the net joint torque doesn't always match the direction of ankle movement. Using the joint angle profile as the reference would allow the assistive torque to match the ankle movement direction. As a result, the natural gait kinematics would not be hindered. The reference ankle joint angle profiles in the sagittal plane (plantarflexion/dorsiflexion) were obtained from a sample of 13 young adults who took part in an earlier study [38] and were asked to walk back and forth on a 12-meter-long walking platform while having their heel strike coinciding with the beat of a metronome. The frequencies of the metronome were set in accordance with the cadence levels of 48 steps/minute, 60 steps/min, and 72 steps/min. An eight-camera motion capture system (VICON, Oxford Metrics, Oxford, UK) was used to collect kinematic data and calculate the ankle joint angle during walking. The reference ankle joint angle profile was calculated by taking the mean profile of the ankle joint angle from 20 randomly selected gait cycles of each young adult. The gait cycle was segmented by the minimum vertical positions of the heel marker at the right foot. Such reference data from younger participants were purposely selected because their ankle profile would be more stable than the target population (i.e., the elderly).

TABLE I
PARTICIPANTS' DEMOGRAPHIC INFORMATION (MEAN ± SD)

	Low risk group	*Medium-high risk group*
Number of participants	*12*	*12*
Number of male participants	*6*	*5*
Age (years):	*64.6±4.1*	*74.2±2.1*
Height (cm):	*163.5±6.4*	*161.9±8.3.0*
Weight (kg):	*64.0±6.9*	*56.7±7.6*
Fall risk assessment score:	*0.5±0.5*	*10.3±2.2*

Within one gait cycle, the actuator had two inflation - deflation cycles. The PAM inflation was initiated to generate assistive torques at the moment when the ankle joint started to dorsiflex. The PAM maintained the inflation status until the ankle joint started to plantarflex at the time of heel-off. Then, the PAM inflated again at the time of toe off to provide assistive torques during the swing phase. The PAM started to deflate after the subsequent heel strike. This process was repeated among gait cycles. Figure 5 shows an example of the control of PAM inflation and deflation in a gait cycle.

B. Experiment

Twenty-four older adults over 60 participated in the experiment. Their fall risks were assessed by the fall risk assessment tool developed by the National Health Commission of the People's Republic of China [39]. The tool can generate an assessment score between 0 to 53. Low, medium and high fall risks correspond to the scores of "0-2", "3-9", and "10 and above", respectively. Table I provides the participants' demographic information and their Fall risk assessment score.

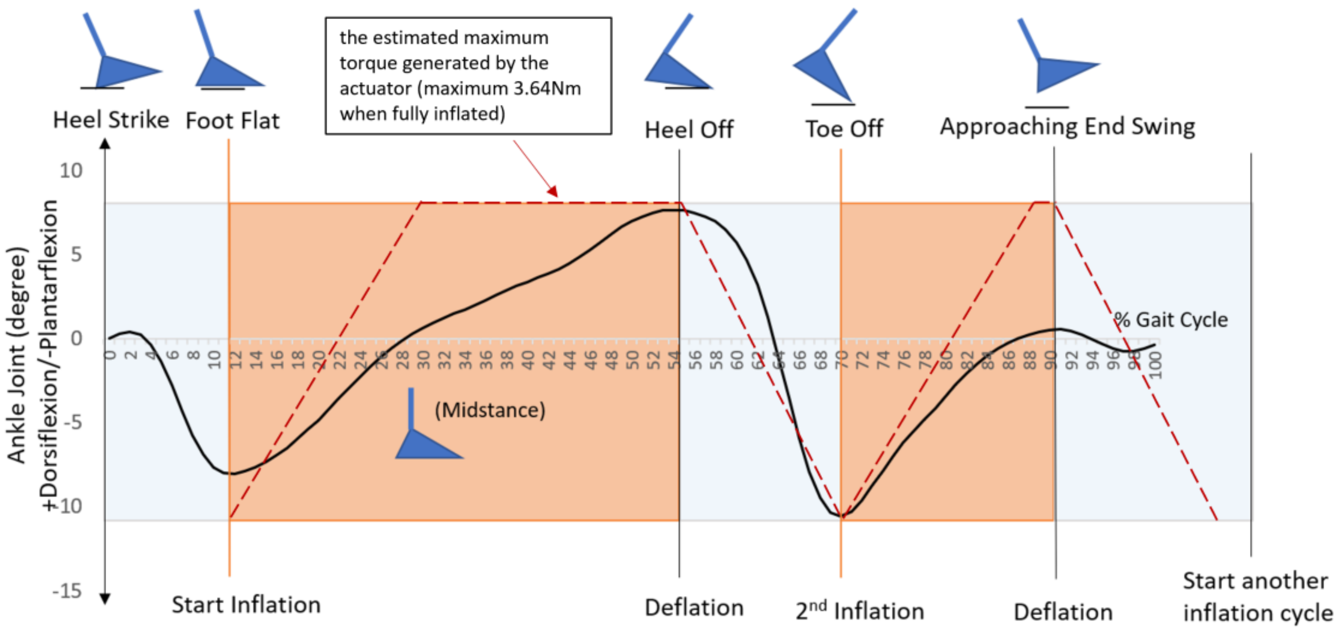

Fig. 5. An example of the control of PAM inflation and deflation in a gait cycle. The red dash line shows the estimated assistive torque profile of the actuator. The maximum assistive torque ~3.64Nm was generated when the PAMs were fully inflated.

It is interesting to note that no participants in the medium-high-fall-risk group had very high fall risk scores (>40). An explanation is that the participants in our experiment were independent as they all came to our lab to complete the experiment on their own. Thus, it is very unlikely for them to have a very large fall risk score. According to the assessment score, 12 participants were classified into the low risk group, and 12 were in the medium-high risk group. Prior to the experiment, written informed consent approved by the local ethics committee was obtained from each participant. This study was approved by the Institutional Review Board of Shenzhen University (Approval No. 20190012).

During data collection, a metronome was used to control the actual cadence of each participant which was set at three different levels: 48 steps/min, 60 steps/min, and 72 steps/min. The participants were asked to walk on an instrumented treadmill (Zebris, Schein, Germany) while having their heel strike in line with the beat of the metronome. The controlled cadence was implemented to minimize the phase-shift between the participant's gait cycle and the pre-programmed inflation-deflation cycle of the soft robotic system, and to allow the soft robotic system to provide the in-time assistance to the participants. The treadmill speed was adjusted according to the preference of each participant under each cadence level, which was determined prior to data collection using a protocol described by Jordan *et al.* [40]. The mean ± SD self-selected preferred speed was 0.57 ± 0.12 m/s, 0.82 ± 0.12 m/s, and 1.09 ± 0.27 m/s, for the cadence levels of 48 steps/min, 60 steps/min, and 72 steps/min, respectively. Each participant went through three testing conditions, i.e., no soft robotic intervention, inactive soft robotic intervention, and active soft robotic intervention. Before data collection, the participants were given as much time as they wanted to get familiar with the device and the experimental set-up. The average time the participants took for familiarization was approximately ten minutes. In the no intervention condition, the participants did not wear the soft robot during walking on the treadmill. In the inactive intervention condition, the participants wore the soft robot but the robot was not activated. In the active intervention condition, the participants wore the soft robot and the robot was activated. The participants completed one 2-min walking trial under each combination of the intervention conditions and cadence levels. The intervention conditions and cadence levels were presented in a random order across the participants to minimize order effects. One-minute break was given in two consecutive trials to avoid potential confounding effects caused by fatigue.

An eight-camera motion capture system (VICON, Oxford Metrics, Oxford, UK) was used to collect kinematic data at the sampling rate of 100 Hz. Sixteen reflective markers were placed on the bony landmarks of the lower body according to VICON's Lower Body Plug-in-Gait Model. Additional eight markers were placed on the left and right shoulders, the clavicula, and sternum. In the inactive and active intervention conditions, four more markers were evenly placed along the lateral side of each actuator for visualization purpose. Figure 6 illustrates the reflective marker placement in the experiment. The output data from the motion capture system were filtered using a fourth order, low pass Butterworth filter with a cut-off frequency of 10 Hz.

C. Data Analysis

Step length and step width were calculated as the anterior-posterior and the medial-lateral distance between the left and right heel markers at sequential heel strikes. The heel strikes were determined by the pressure sensors embedded under the treadmill, when the vertical ground reaction force estimated by the pressure sensors first exceeded 5N. Data from the 11th gait cycle to the 40th gait cycle in each walking trial were used

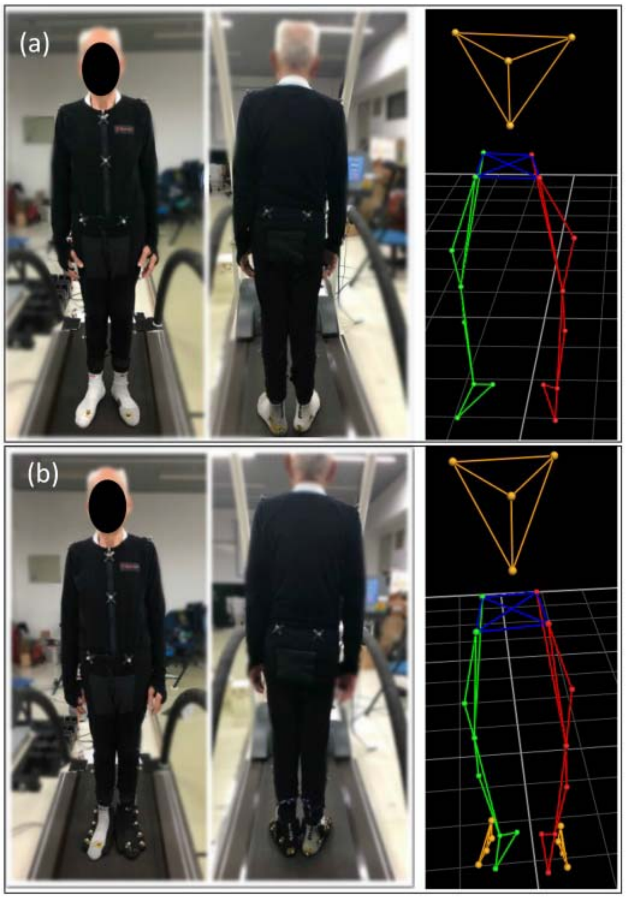

Fig. 6. The reflective marker placement in the experiment. (a) without intervention; (b) with intervention.

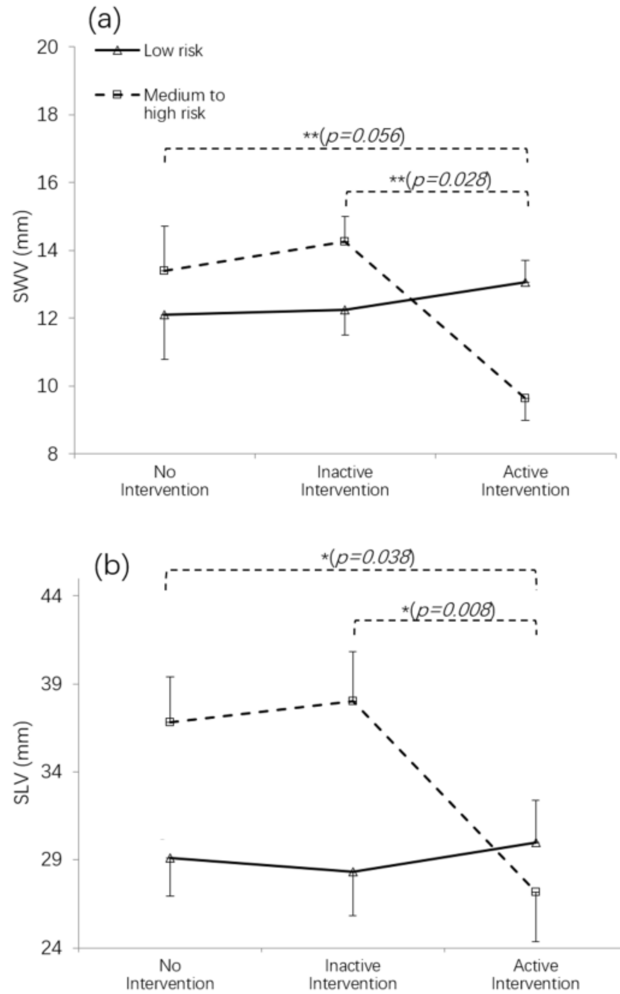

Fig. 7. Interaction effects between fall risks and intervention. (a) step width variability (SWV) at 48 s/m; (b) stride length variability (SLV) at 60 s/m.

to calculate gait variability. Gait variability was quantified by the standard deviations of step length and step width using the method suggested in Qu [41].

The normality and sphericity of the dependent variables were confirmed by Shapiro-Wilk test and Mauchly's test, respectively. The test results indicated that the assumptions of normality and sphericity had not been violated. A mixed-model ANOVA with two categorical independent variables (i.e., fall risks and intervention, with a random effect of participants) was carried out to investigate the effects of fall risks (low versus medium-high risk) and intervention (no intervention versus inactive intervention versus active intervention) on step length variability and step width variability at each cadence level. Post-hoc pairwise comparisons were conducted when necessary by using the Bonferroni correction. Where significant interaction was found between 'fall risks' and 'intervention', one-way ANOVA was carried out for the low risk group and medium-high risk group separately with the 'intervention' as the independent variable. The level of significance was set at 0.05.

III. Results

Older adults with medium-high fall risk had significantly larger step length variability at the cadence level of 72 steps/min when compared with the low-risk older adults (Table II). Active intervention was observed to significantly reduce step length variability at the cadence levels of 48 steps/min and 72 steps/min, but inactive intervention did not make any differences in step length variability (Table II and Table III). Step width variability in the intervention conditions was not significantly different from that in the no intervention condition. However, when making comparisons between intervention conditions, inactive intervention had significantly larger step width variability at the cadence level of 48 steps/min than did active intervention (Table II and Table III).

Significant interaction effects were found in step-width variability at 48 steps/min and step length variability at 60 steps/min (Table II). One-way ANOVA results showed that active intervention was effective in reducing step width variability at 48 steps/min and stride length variability at 60 steps/min only in the medium-high risk group (Figure 7). Figure 8 shows the mean profiles of the lower-limb joint angles (hip, knee, and ankle) under different intervention conditions, with each row representing the data under different cadence levels (from the top to the bottom row: 48 steps/min, 60 steps/min, 72 steps/min, respectively).

IV. Discussion

This study presented a novel soft robotic intervention for gait enhancement among older adults. The intervention was an ankle-based soft robotic system that was developed to

TABLE II

RESULTS FROM TWO-WAY ANOVAS (MEAN ± SD). SLV = STEP LENGTH VARIABILITY; SWV = STEP WIDTH VARIABILITY; M-H = MEDIUM-HIGH

	Fall Risk			*Intervention (two groups combined)*				*Interaction Effects*
	Low risk	*M-H risk*	*p-values*	*No*	*Inactive*	*Active*	*p-values*	*p-values*
SLV at 48 steps/min (mm)	*29.8±7.8*	*32.9±11.9*	*0.315*	*34.9±13.8*	*30.0±7.1*	*28.3±5.1*	*0.035**	*0.161*
SWV at 48 steps/min (mm)	*12.5±3.1*	*12.4±3.9*	*0.964*	*12.8±4.6*	*13.3±2.7*	*11.4±2.8*	*0.041**	*0.001**
SLV at 60 steps/min (mm)	*29.1±8.1*	*34.0±9.2*	*0.096*	*32.4±8.5*	*32.5±9.6*	*28.8±8.4*	*0.050*	*0.009**
SWV at 60 steps/min (mm)	*13.8±4.1*	*13.7±4.4*	*0.938*	*13.4±4.4*	*14.6±4.9*	*13.1±3.3*	*0.245*	*0.108*
SLV at 72 steps/min (mm)	*27.7±7.8*	*35.2±10.6*	*0.040**	*34.2±10.5*	*30.2±11.2*	*29.3±7.3*	*0.028**	*0.149*
SWV at 72 steps/min (mm)	*14.3±3.8*	*15.8±4.8*	*0.31*	*14.6±4.2*	*15.9±4.8*	*14.6±3.3*	*0.200*	*0.344*

* indicates statistical significance (p-value<0.05).

TABLE III

RESULTS FROM POST-HOC MULTIPLE COMPARISONS. SLV = STEP LENGTH VARIABILITY; SWV = STEP WIDTH VARIABILITY; CI = CONFIDENCE INTERVAL."(I)" AND "(J)" REPRESENT THE INTERVENTION CONDITIONS IN THE SAME COLUMN. "(I)-(J)" SUGGESTS THE DIFFERENCES OF THE DEPENDENT VARIABLES BETWEEN THE INTERVENTION CONDITIONS OF (I) AND (J)

	Intervention		*(I)-(J)*	*p-values*	*95% CIs*	
	(I)	*(J)*				
SLV at 48 steps/min (mm)	*No*	*Inactive*	*5.193*	*0.384*	*-3.397*	*13.783*
	No	*Active*	*7.609*	*0.037**	*0.387*	*14.83*
	Inactive	*Active*	*2.416*	*0.502*	*-2.016*	*6.847*
SWV at 48 steps/min (mm)	*No*	*Inactive*	*-0.503*	*~1.00*	*-2.849*	*1.844*
	No	*Active*	*1.399*	*0.186*	*-0.444*	*3.243*
	Inactive	*Active*	*1.902*	*0.015**	*0.324*	*3.48*
SLV at 72 steps/min (mm)	*No*	*Inactive*	*3.88*	*0.249*	*-1.687*	*9.447*
	No	*Active*	*4.949*	*0.022**	*0.629*	*9.27*
	Inactive	*Active*	*1.07*	*~1.00*	*-3.616*	*5.755*

* indicates statistical significance (p-value<0.05).

provide assistive torque for ankle dorsiflexion during walking. The results showed that when being activated, the soft robotic intervention can result in reduced step length variability among older adults, suggesting that it has positive effects on controlling gait variability. As increased gait variability was observed to be associated with increased fall risks [5], [6], the proposed intervention could possibly serve as a potential solution to fall prevention among older adults.

It was also found that gait variability was not affected by the inactive intervention when compared to the no intervention condition. The difference of the active and inactive interventions mainly lies in whether the intervention generates assistive torque or not. Thus, the observed positive active intervention effects on step length variability can be attributed to the assistive torque generated to facilitate ankle dorsiflexion. Age-related declines in dorsiflexor strength have been reported, and such declines were more substantial in fallers versus non fallers [42]. Kang and Dingwell [43] presented that decreased dorsiflexor strength was a major factor contributing to increased gait variability for older adults. When our proposed intervention was activated, assistive torque was generated to facilitate ankle dorsiflexion during both midstance phase and swing phase (Figure 4). Such assistive torque could counteract the effect of declined dorsiflexor strength, and thus reduce step length variability for older adults.

When examining the interaction effects of intervention and fall risks, it is interesting to know that active soft robotic intervention could help reduce gait variability for older adults with medium-high fall risks, but did not have effects for older adult with low risks. This finding suggests that the active intervention might be more effective for people with higher fall risks. This is probably because the older adults with higher fall risks might have decreased dorsiflexor strength compared to those with lower fall risks [43]. Thus, they are more likely to benefit from this soft robotic intervention since it is designed to provide assistive torque for ankle dorsiflexion.

It was found that older adults with medium-high fall risks had larger step length variability than those with low fall risks at cadence level of 72 steps/min. This finding is consistent with previous studies [25], and further supports that increased gait variability could be associated with increased fall risks. It was also worth noting that between-group differences did not exist in step length variability at lower cadence levels (i.e., 48 steps/min and 60 steps/min). Researchers have reported that slower walking speed leads to increases in postural stability [44]. In addition, interaction results also presented that active intervention led to greater gait variability reduction at lower cadence levels in the medium-high risk group versus the low risk group. These all suggest that when walking at lower cadence levels, older adults with medium-high fall risks would have improved postural stability and reduced fall risks. As a result, their gait variability was not significantly different from that of the low risk participants.

The intervention effects on step width variability were also investigated because step width variability might be a more meaningful descriptor of locomotion control than step length variability for elderly [45]. However, the results suggested that this intervention could not improve step width variability. This can be explained by the fact that the assistive torque was designed for dorsiflexion which mainly controls the movement in the sagittal plane. It would be of interest to develop a soft robotic intervention that can generate assistive torques for inversion and eversion, and evaluate its effects on step with variability in future research.

The proposed soft robotic intervention has several attractive features. Specifically, the PAMs were used as the actuators, which made the presenting system superior to previously reported powered ankle-foot orthoses based on rigid structures and motor-driven actuators [46] in terms of comfort and safety. We chose the PAMs proposed by Mosadegh *et al.* [27] due to its good weight-to-torque ratio and fast actuating property. To our best knowledge, this is the first time for this type of PAM being used in a gait enhancement application. The selected PAMs are compliant and lightweight. The additional mass attached to the ankle was only 76.6g. Doing so allows minimal interference with users' natural gait kinematics and ensures safe and comfortable interaction between users and the intervention. Besides, no obvious deviations can be observed for the low limb gait kinematics across the intervention conditions (Figure 8). This provided empirical evidence for minimal interference of the proposed intervention with the lower-limb joint kinematics. This was not to our surprise, given that assistive torque (i.e., $\sim 3.64 \pm 0.27$ Nm) generated by the proposed intervention was small. In addition, findings from the present study showed that our proposed soft robotic intervention was able to help control gait variability. Thus, it can benefit older adults by reducing their fall risks.

Some PAMs, particularly the McKibben type, have been previously used in powered ankle-foot orthoses. For example, Park *et al.* [47] proposed an ankle-based soft robotic device. Four McKibben PAMs were placed on the lower leg to provide assistive forces that can facilitate ankle dorsiflexion, plantarflexion, inversion, and eversion for people with pathological gait. However, their prototype was not tested for dynamic gait. Another example is given by Ferris *et al.* [48], [49] who designed a pneumatically powered AFO using two McKibben PAMs attached both anteriorly and posteriorly to mimic the dorsiflexor and plantarflexor. However, this AFO still consisted of rigid parts such as the carbon fiber shank section and steel hinge joint. Our soft robot has better wearability with all the wearable components made by soft materials. Furthermore, unlike previously reported soft robotic device or powered orthosis that typically used linear actuators [21], [46]–[49], the proposed intervention adopts a bending mechanism of the PAMs to provide direct assistive angular torque to the ankle joint when being activated. The proposed intervention is more efficient in terms of force transmission.

The proposed intervention system was tethered to air compressors. However, its operation air pressure level (110 kPa) was lower compared to previously reported wearable robotic systems with pneumatical actuators [25]–[27]. Thus, it is feasible to implement miniature and portable air compressors in our system, which could be comfortably worn and make minimal interference with the activities of daily living. This can allow our system to be used as a daily assistive device.

The PAMs used in this study can be inflated and deflated within 250 ms. The results showed that such inflation/deflation speed could be sufficient for the intervention in supporting and assisting slow gait. The cadence levels tested here were below the mean cadence level during daily activities of the elderly. Previous studies suggested slower walking speed can result in increased gait variability for the elderly by creating additional constraints on their neuromuscular apparatus [34]. Additionally, slow walking speed was also found to have effects on stride length [50]. Thus, the findings from the present study may only be applicable for slow walking conditions. Future studies will be focused on improving inflation and deflation efficiency of PAMs and investigating the interactions between the soft robotic intervention and walking speed quantitatively.

During the experiment, the participants were given approximately ten minutes to get familiar with the device. The effects of the soft robotic intervention were tested immediately after the familiarization procedure. It is possible that more training or adaptation time could make a difference in the intervention effects. Thus, an interesting topic for future research is investigating the long-term effects of the proposed intervention. Furthermore, the ground reaction forces were not collected. Thus, the effects of this system on the biological joint torques were not investigated. This will be one direction for future research. Due to the lack of sensing components, the control strategy adopted for the soft robotic intervention was pre-programmed. This indicates that our proposed intervention solution cannot address individual differences in gait patterns. Though this is a common practice in current relevant studies [23], [51], an individualized solution would lead to better gait enhancement outcome. Therefore, our next step research will focus on developing an individualized real-time control strategy by integrating sensing components into the soft robotic intervention system. For example, miniature IMUs can possibly be mounted on the surface of the foot and the

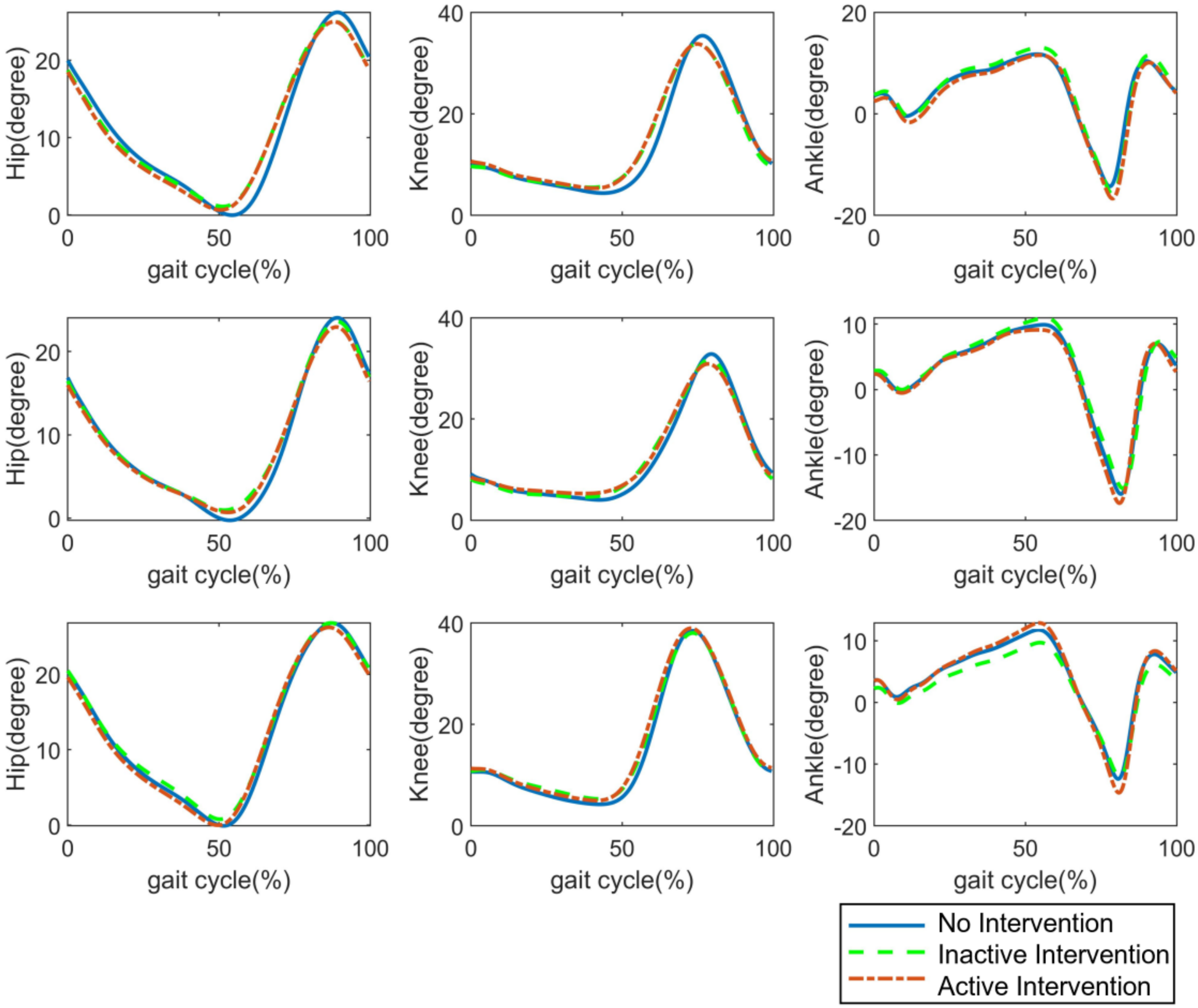

Fig. 8. The mean profiles of the lower-limb joint angles in the sagittal plane under different intervention conditions. Top row: joint angles under cadence level of 48 steps/minute; middle row: joint angles cadence level of 60 steps/min; bottom row: joint angles under cadence level of 72 steps/min.

frontal side of the shank to obtain ankle kinematics (e.g. [52]). With high sampling rate (500Hz to 1000 Hz) of the IMUs, the key temporal events, such as the time when the ankle joint starts to dorsiflex, can be detected with minimal time delays. Such information can be fed to the microcontrollers as the triggering signal to realize the real-time control of the soft robotic intervention system.

V. Conclusion

In summary, findings from the present study showed that when being activated, our proposed soft robotic intervention was able to help control gait variability. Thus, it could be useful for preventing falls among older adults. The major contribution of this study lies in the finding that ankle based soft robotic can be implemented as an affective intervention to enhance gait performance for older adults, especially for those who had medium to high fall risk. By comparing the effects of active and inactive interventions, observed positive active intervention effects can be attributed to the assistive torque generated to facilitate ankle dorsiflexion. This finding implies the importance of incorporating assistive torque for dorsiflexion in fall prevention interventions.

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