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A Novel Approach to Apply Gait Synchronized External Forces on the Pelvis using A-TPAD to Reduce Walking Effort

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Abstract

In this paper, we develop an intervention to apply external gait synchronized forces on the pelvis to reduce the user's effort during walking. A cable-driven robot was used to apply the external forces and an adaptive frequency oscillator scheme was developed to adapt the timing of force actuation to the gait frequency during walking. The external forces were directed in the sagittal plane to assist the trailing leg during the forward propulsion and vertical deceleration of the pelvis during the gait cycle. A pilot experiment with five healthy subjects was conducted. The results showed that the subjects applied lower ground reaction forces in the vertical and anterior-posterior directions during the late stance phase. In summary, the current work provides a novel approach to study the role of external pelvic forces in altering the walking effort. These studies can provide better understanding for designing exoskeletons and prosthetic devices to reduce the overall walking effort.

Index Terms

Physically Assistive Devices; Human Performance Augmentation; Tendon/Wire Mechanism

I. Introduction

Walking is the main mode of locomotion among legged animals. Humans accomplish walking through successful inter-limb coordination and cyclic activation of leg muscles. Human walking is a mechanically-complex task that involves a state of continuous imbalance, where each step is taken to regain the body's balance [1]. In particular, sufficient foot clearance is required when one leg is in the swing phase to prevent scuffing with the ground. At the end of the swing phase, the leg must be positioned ahead of body. This is accomplished through bearing weight on the leg in contact with ground (stance phase). At the end of a leg's stance phase, sufficient push-off force is needed to initiate the swing phase of that leg as well as to propel the body's center of mass (COM) forward.

Given the ease with which healthy individuals accomplish these patterns, researchers have described the steady-state walking as the most metabolically economic locomotion mode [1]–[5]. Despite this, a significant amount of research effort has explored ways to augment the metabolic cost of walking to make walking easier. The main motivation for these efforts

is to restore performance in individuals with lower limb impairments, such as lower limb amputees, and to enable increased human endurance while performing demanding tasks while walking, such as carrying weight. To achieve these goals, researchers have developed external devices with passive and active force generating elements to complement the functioning of lower limb during walking.

In the literature, simple inverted pendulum walking models have been developed to understand the human walking dynamics [2]–[4]. During walking leg motions are coordinated to move the body's COM forward while keeping at least one foot in contact with the ground at all times, thereby inducing an arc shaped trajectory to the COM in the sagittal plane. In these mathematical models, step to step transition has been identified as the most critical event. In this phase, the trailing leg performs positive work to smoothly change the COM motion in the vertical and anterior-posterior directions. Studies on lower limb joint moments identify ankle and hip joints as the source of the positive work with ankle being the dominant contributor [4], [6].

The majority of the works reported on reducing the walking effort have thus focused on developing an exoskeleton or prosthetic device to complement the ankle joint, [5]–[13]. The exoskeletons have been tested on able bodied individuals to study the changes in the metabolic cost under altered walking conditions, such as walking on an incline. Similarly, the prosthetic devices have been tested with amputees to study changes in the metabolic cost and coordination between the intact and affected leg. In the literature, researchers have also developed mechanisms to study the role of applying external forces on the thigh segments to complement the hip joint during walking [14], [15].

Among these works, very few studies have been successful in reducing the overall metabolic cost of walking compared to those with able bodies [5] or individuals using conventional prosthesis [7]. One important factor in using these exoskeletons or prosthetic devices is that they change the human dynamics, in particular due to the added mass on the leg and mismatch in timing when the external forces are applied [6], [12]. Since humans adapt their walking pattern under different external conditions, it becomes critical to understand the adapted pattern.

As the main goal of inter-limb coordination and cyclic activation of leg muscles during steady-state walking is to propel the COM forward, studies with external forces that directly alter the COM dynamics can provide better understanding for designing exoskeletons and prosthetic devices to reduce the overall walking effort. *In this work, we present a novel methodology to apply gait synchronized external forces at the pelvis. A cable-driven active tethered pelvic assist device, A-TPAD* (Fig. 1(a)), was used to apply the external forces and an adaptive frequency oscillator (AFO) scheme was developed to adapt the applied force actuation timing to a subject's walking pattern. An experiment with five healthy subjects was conducted, where external forces were applied at the pelvis to assist the trailing leg during the stance phase. It was hypothesized that the subjects will reduce the trailing leg's effort during walking in the presence of applied force.

II. Method

A. Pelvic Intervention using A-TPAD

Due to the proximity of the pelvis segment to the body's COM, pelvic motion plays a central role in all forms of locomotion. The vertical pelvic motion and acceleration profiles during steady-state walking are presented for a healthy individual in Fig. 1(b) over a gait cycle [16]. It is observed that human pelvis undergoes two oscillations along the vertical axis. Each oscillation is accompanied with periodic exchange between the gravitational potential and kinetic energy.

The transition in the pelvic motion direction during the single support (SS) phase is mainly due to the external gravitational force on the body. While the motion transition during the double support (DS) phase is to push the pelvis into the periodic arc shaped trajectory, which is achieved through successful step to step transition. Thus, a significant amount of lower limb muscle effort is needed from the trailing leg to provide forward propulsion during the DS phase and to decelerate the vertical fall of the COM during the SS phase [3], [4], [6], [17].

In this study, an external force vector was applied at the pelvis to reduce the trailing leg effort over a gait cycle. A four cables A-TPAD configuration was used to apply the external wrench to the pelvis (Fig. 1(a)), where two cables connect the hip belt to the top of the frame and two cables connect the hip belt to the bottom of the frame. By controlling the tension magnitude in these cables, an external force vector was applied to the pelvis. The design and control strategy of the A-TPAD were described earlier in [18].

B. Desired External Wrench

In the current study, the sagittal plane component of the force vector (F_Y : anterior-posterior, and F_Z : vertical) were controlled to provide the external assistance during walking, which are described as

$$\begin{aligned} F_Y &= F_{YC} + F_{YV} \\ F_Z &= F_{ZC} + F_{ZV} \end{aligned} \quad (1)$$

where F_{YC} and F_{ZC} are the constant terms, while the F_{YV} and F_{ZV} are the varying terms of the force vector that appear briefly during a part of the gait cycle (Fig. 1(c)). During the experiment, the remaining force and moment components were maintained in a small range to meet the workspace requirement. In particular, the medial-lateral force component, F_X , was maintained between $\pm 1\%$ of a subject's body weight (BW), and the moment components resolved at the pelvic center were maintained between ± 4 Nm.

Anterior-posterior force component (F_Y)—The varying term F_{YV} was applied to the pelvis during the DS phase of a gait cycle. It was activated at 1% of a gait cycle (GC) past a heel-strike event. The right and left leg gait cycles (RGC and LGC) were identified separately using an adaptive frequency oscillator (Section II-C) to adapt the timing of force

actuation to the gait frequency during walking. Once activated, the F_{YV} magnitude grew as a sinusoidal function of time from zero to a maximum value of 2% BW. The F_{YV} term was deactivated exactly at the end of the double support phase, i.e., at the toe-off event which was identified using a foot switch in each shoe. Thus, there were two instances of F_{YV} during a gait cycle: from 1% RGC to LTO and from 1% LGC to RTO as shown in Fig. 1(c). The choice of this profile was made to mimic the biological forward propulsion effort applied by the lower limb muscles of the trailing leg, which starts early in the double support phase to generate sufficient propulsion force to accomplish the weight transition and to propel the body's COM forward at push-off [4], [12], [17]. *It was therefore expected that the applied force intervention will reduce the trailing leg's walking effort required for the forward propulsion during the late stance phase.*

Vertical force component (F_Z)—The varying term F_{ZV} was activated at 30% GC past the heel-strike event of a leg, and deactivated at 5% GC past the heel strike event of contra-lateral leg. In this time span, the magnitude of F_{ZV} grew as a sinusoidal function of time from zero to reach a maximum value of 3% BW which was maintained till it was deactivated. Thus, the F_{ZV} term was applied twice during a gait cycle: from 30% RGC to 5% LGC and from 30% LGC to 5% RGC as shown in Fig. 1(c). Similar to the F_{YV} profile, the choice of F_{ZV} profile was motivated from the biological walking effort spent by the leg in contact with the ground in decelerating the body's COM during the single support phase. As the COM reaches its vertical peak position almost around mid single support phase and starts falling under the gravitational force afterward, a significant trailing leg's muscle activation is required for deceleration to prevent sudden collision and accomplish foot transition [4], [13]. *It was therefore expected that the applied force intervention will further help in reducing the trailing leg's walking effort during the late stance phase.*

As the F_{YV} and F_{ZV} values were zero for most part of the gait cycle, a constant term F_{YC} with 2% BW magnitude and F_{ZC} with magnitude 4% BW were maintained to keep the cables in tension.

C. Gait Phase Detection

Algorithm 1

Gait Phase Detection

-
- 1: Input: $M \leftarrow 2$; $v \leftarrow 2$; $e \leftarrow 20$; $f_{min} \leftarrow 1.3$ Hz
 - 2:

$$\text{Zero values: } Y_0 \leftarrow \begin{bmatrix} \phi_0 \\ \omega_0 \\ \alpha_0 \\ \beta_0 \end{bmatrix} = \begin{bmatrix} 0_{M \times 1} \\ 2\pi \times f_{min} \\ 0_{M \times 1} \\ 0_{1 \times 1} \end{bmatrix}_{(2M+2) \times 1}$$
 - 3: Acquire: $a_y \leftarrow$ pelvic vertical acceleration
 - 4: $\theta_d \leftarrow (a_y - \text{acceleration due to gravity})$
 - 5:

$$\hat{\theta}_t \leftarrow (\beta_{t-1} + \sum_{i=1}^M \alpha_{t-1}(i) \sin(\phi_{t-1}(i)))$$
 - 6: $R(t) \leftarrow (\theta_d - \hat{\theta}_t)$ ▷ Teaching signal

7:
$$\begin{bmatrix} \dot{Y}_t(i) \\ \dot{Y}_t(M+1) \\ \dot{Y}_t(M+1+i) \\ \dot{Y}_t(2M+2) \end{bmatrix} \leftarrow \begin{bmatrix} \dot{\phi}_t \\ \dot{\omega}_t \\ \dot{\alpha}_t \\ \dot{\beta}_t \end{bmatrix} = \begin{bmatrix} iY_{t-1}(M+1) + \varepsilon F(t)\cos(Y_{t-1}(i)) \\ \varepsilon F(t)\cos(Y_{t-1}(1)) \\ \nu F(t)\sin(Y_{t-1}(i)) \\ \nu F(t) \end{bmatrix}$$

8: $Y_t \leftarrow \int \dot{Y}_t$

9: $Y_{\lambda}(1) \leftarrow \text{mod}(Y_{\lambda}(1), 4\pi)$ $\triangleright$ Remainder after division

10: **if** $Y_{\lambda}(M+1) < 2\pi f_{min}$ **then**

11: $Y_{\lambda}(M+1) \leftarrow 2\pi f_{min}$

12: **return** $Y_{\lambda}(1)$ and $Y_{\lambda}(M+1)$ $\triangleright$ Gait phase and frequency

An adaptive frequency oscillator (AFO, [19]) scheme was implemented to detect the gait phase during the experiment, the methodology is presented in Algorithm 1. An accelerometer was placed on the subject's pelvis during the experiment to record the vertical pelvic acceleration values, a_y . An estimated acceleration signal, $\hat{\theta}$, was constructed using M harmonic oscillators and the previous estimated value, as described in line 5 of Algorithm 1. The amplitude, α , frequency, ω , offset, β , and phase, ϕ , of these harmonic oscillators describe the AFO states, Y (line 2). A teaching signal, $F(t)$, was constructed from the vertical pelvic acceleration values (minus the acceleration due to gravity), θ_d , and the $\hat{\theta}$ values (line 6). The first order equations, $\dot{Y}_t$ in line 7, were solved to learn the frequency of $F(t)$. Here, ε , ν , and f_{min} are the coupling strength, learning factor, and minimum frequency limit for the AFO respectively. At a time instant, the phase of the first harmonic oscillator, $\phi(1)$, approximated the gait phase value [20]. The gait phase detection algorithm was tested for different walking speeds and a good estimate of gait frequency was obtained using $M=2$, $\varepsilon=20$, $\nu=2$, and $f_{min}=1.3$. With these values, sudden changes in the gait frequency were adapted within three walking cycles. In Fig. 2(a), the θ_d and corresponding $\hat{\theta}$ data are shown for a section of the experiment.

D. Wrench Planner

As pelvis oscillates vertically at twice the gait frequency, the gait phase, ϕ , values calculated from Algorithm 1 vary from 0 to 4π over a gait cycle. To estimate the percentage of gait cycle with respect to each leg, ϕ_{RGC} and ϕ_{LGC} , force sensitive resistors based foot switches were used to detect the heel-strike, HS, and toe-off, TO, events. The ϕ values at each HS event, ϕ_{RHS} and ϕ_{LHS} , were used to compute the ϕ_{RGC} and ϕ_{LGC} values, as described in lines 3 and 4 of Algorithm 2. In Fig. 2(b), the foot switch data from both legs along with the ϕ_{RGC} and ϕ_{LGC} values are presented for a section of the experiment. It is observed that the ϕ_{RGC} and ϕ_{LGC} vary from 0% at the HS event to 100% at the following HS event. During the experiment, the varying vertical force component, F_{ZV} , was applied from 30% ϕ_{RGC} to 5% ϕ_{LGC} and from 30% ϕ_{LGC} to 5% ϕ_{RGC} (lines 5–8). Similarly, the varying A-P force component, F_{YV} , was applied from 1% ϕ_{RGC} to LTO (left leg toe-off) and from 1% ϕ_{LGC} to RTO (lines 9–12). The constant force terms F_{ZC} and F_{YC} were applied throughout the training session.

Algorithm 2

Wrench Planner

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1: Available:  $\phi$  and  $\omega$                                 ▷ Gait phase and frequency
2: Acquire: HS & TO  $\leftarrow$  foot switches
3:                                                    ▷ Right gait phase
    $\phi_{RGC} \leftarrow \frac{100 \times (\phi - \phi_{RHS})}{4\pi}$ 
4:                                                    ▷ Left gait phase
    $\phi_{LGC} \leftarrow \frac{100 \times (\phi - \phi_{LHS})}{4\pi}$ 
5: if ( $\phi_{RGC}$  30% &&  $\phi_{LGC}$  5% || ( $\phi_{LGC}$  30% &&  $\phi_{RGC}$  5%)) then
6:
    $\text{return } F_Z \leftarrow (F_{ZC} + \overbrace{A_Z \sin(\omega t)}^{F_{ZV}})$ 
7: else
8:    $\text{return } F_Z \leftarrow F_{ZC}$ 
9: if (1%  $\phi_{RGC}$  && LTO) || (1%  $\phi_{LGC}$  && RTO) then
10:
    $\text{return } F_Y \leftarrow (F_{YC} + \overbrace{A_Y \sin(\omega t)}^{F_{YV}})$ 
11: else
12:    $\text{return } F_Y \leftarrow F_{YC}$ 

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III. Human Experiment

A. Protocol

Five healthy subjects, three male and two female in age range 22–30 years (mean age: 26.6 yrs) and mean weight 67.6 kg (SD: 9 kg), participated in the experiment. The training protocol was approved by the Columbia University Internal Review Board and involved a Baseline (BL) and Training (T) sessions. During both BL and T, each subject walked for twelve minutes at a constant speed of 3.8 kmph. Data were collected for a minute four times during each session, at 1st, 4th, 7th, and 10th min. Cables were attached to the hip belt for the Training session to apply the desired external wrench to the pelvis. A rest brake of ten minutes was given to each subject between the BL and T sessions.

Each subject was suited up with retro-reflective markers to record the human motion data [21]. Surface EMG activities from the lower limb muscles were measured bilaterally, namely Gastrocnemius (GAS), Soleus (SOL), and Tibialis Anterior (TA). A fabric hip belt with cable attachment points was worn by the subject, and a three axis accelerometer was mounted on the subject's pelvis. Force sensitive resistor (FSR) pressure pads were mounted on the subject's shoe insoles to measure the foot pressure data. An Oxycon metabolic system from Care Fusion (CA, USA) was used to measure breath by breath volumetric oxygen (VO₂) intake. A heart rate belt from Polar (NY, USA) was used to obtain the breath by breath heart rate (HR) values. In addition, a split-belt treadmill with force plates from Bertec

(Columbus, OH) was used to measure the ground reaction forces (GRF) for each leg during the experiment.

B. Data Processing

During the experiment, each subject's EMG, GRF, VO₂ intake, HR, and applied wrench data were recorded for the analysis. The time histories of all gait parameters were normalized in time to 100% of the gait cycle. The GRF data were recorded at 1000 Hz and post-processed using a low-pass filter (4th order Butterworth, cut-off 20 Hz). EMG signals were post-processed using a band-pass filter (4th order Butterworth, 40–450 Hz) and a full-wave rectification. The signals were smoothened using a low pass filter ($n=4$, $f_c = 6$ Hz). For every subject, EMG data of each muscle were normalized to the peak values recorded during the baseline session. The metabolic system data was normalized with subjects' body weight to get metabolic consumption rate (VO₂/kg). First two minutes of VO₂ data collected during the BL and T sessions were discarded due to increased physiological demand of oxygen by the muscles. The remaining ten minutes of data were filtered using a 4th order low pass filter at a cut-off frequency of .04 Hz [22]. The first minute, last minute, and overall ten minutes average VO₂/kg were calculated for each session per subject. For the HR data, a 21-sample block average was taken of the raw data. The smoothed HR data were averaged for each session subjectwise.

To study the performance and response of the group to the applied external wrench, gait measures from the last five gait cycles for trials BL1, BL4, and T4 and from the first five gait cycles for trial T1 were analyzed. The data were first checked against the sphericity violation using the Mauchly's test and the Huynh-Feldt correction was applied when the data violated the condition. One way repeated measure ANOVA was performed to determine the statistical significance (defined as $p < 0.05$). The Bonferroni-Holm significant difference test was performed when a statistical significance was identified. The chosen combinations for statistical difference comparison were BL4-T1 and BL4-T4. For the VO₂ and HR parameters, a pairwise t-test assuming unequal variance was performed between BL and T (significance defined at $p < 0.05$).

C. Results

The right leg ground reaction force components normalized with the subject's body weight in the anterior-posterior (GRF_Y) and vertical (GRF_Z) directions are shown in Fig. 3(a) for a representative subject during different trials. Distinctive differences can be observed in the GRF values during the BL and T trials. In the anterior-posterior direction, the GRF_Y values during the T1 and T4 trials were higher than the baseline values in the early stance phase but were lower in the late stance phase. In the vertical direction, lower GRF_Z values were observed during the training trials around the late stance phase on each side.

Average GRF values of right leg for the group are presented in Fig. 3(b). In the anterior-posterior direction, the GRF_Y propulsive impulse values (time integral of negative GRF_Y values) reduced during the training trials. The changes were significant for the right leg ($p < 0.05$), and significant differences were reported between BL4-T1 and BL4-T4. In addition, the GRF_Y braking impulse values (time integral of positive GRF_Y values) increased

significantly for both left and right legs ($p < 0.05$). Pairwise comparison showed significant changes between BL4-T1 and BL4-T4. Notably, the GRF_Z values reduced for both legs during the training session. The statistical analysis reported significant reduction in the GRF_Z peak and impulse (time integral of GRF_Z values over a gait cycle) values between BL4-T1 and BL4-T4 pairs for both legs ($p < 0.05$).

The weight normalized volumetric oxygen consumption rate (VO_2/kg) values are presented in Fig. 4 for each subject over the ten minutes of the BL and T sessions. Lower VO_2 consumption values were observed for subjects 1, 2, and 5 during the training session. A linear fit model was used for the VO_2/kg data during a session as shown in Fig. 4. A decreasing VO_2 consumption rate was observed for all subjects as the training with the external forces progressed. The average VO_2/kg values for each subject during the first minute, last minute, and over the full ten minutes are presented in Table I. The group average of these values during the training session were lower than the baseline values. In addition, the group average of HR values also reduced during the training session. In particular, VO_2 consumption rate and HR values for subjects 1, 2, and 5 decreased in presence of external forces but increased for subjects 3 and 4. Due to these differences in subjects' responses to the force training statistical significance was not observed when the group data were compared between the baseline and training sessions.

The right leg EMG envelopes are plotted for a represented subject in Fig. 5(a). Lower SOL activities were reported during the training trials, T1 and T4, compared to the baseline activation levels. In addition, the activation level of MG reduced during the training trials. However, the TA activities during the training trials T1 and T4 were observed to be higher than the baseline activation levels. The peak activation level during the major burst of a muscle and the RMS values of the lower limb muscles are presented in Fig. 5(b) over the group. Lower peak and RMS values were observed during the training trials for the SOL and MG of each leg. The statistical analysis reported significance for left leg SOL RMS values and right leg MG peak values ($p < 0.05$). The post-hoc pairwise comparison reported significant difference only between BL4-T4 trials for left leg SOL RMS values. No statistical significance was reported for the changes in the TA peak and RMS values over the group.

IV. Discussion

The goal of the current work was to develop a novel methodology to apply gait synchronized external forces at the pelvis. In the literature [23]–[26], various rigid-link devices have been presented to intervene at the pelvic level. In this study, a cable driven active tethered pelvic assist device, A-TPAD, was used. The A-TPAD has the advantage of using cables which do not add external mass/inertia on the user. In addition, the A-TPAD can be easily customized to use different cable routing configurations to apply a desired force profile [27]–[29]. In comparison, existing rigid-link pelvic devices do not provide control of all six DOFs and thus add undesired human-robot joint misalignment and mobility constraints during walking. In this work, an adaptive frequency oscillator, AFO, scheme was also developed to synchronize the external force actuation timing with lower limb movements and to adapt the force duration to sudden changes in the gait frequency. The use of AFO schemes have been

reliably tested in the literature to determine the gait phase [20], [30]. Notably, in the current work such scheme is used to plan the cable tension values of a cable-driven robot to exert external forces at the pelvis.

In the current experiment, external force field was motivated to assist the trailing leg during walking. The activation timing and duration of the vertical force component were planned to assist the trailing leg during the weight transition on the leading leg by decelerating the vertical fall of the pelvis [4], [13]. Thus, the force was applied during the latter part of the single support phase. The selection of 30% of gait cycle as the activation time was to enforce sufficient upward assist at pelvis just before the contra-lateral leg's heel strike. Similarly, the activation timing and duration of the anterior-posterior force component were planned to assist the trailing leg during the forward propulsion at push-off [4], [12], [17]. Thus, the force was applied during the double support phase, starting at 1% of gait cycle past the HS and terminating at the toe-off (TO). Notably, reductions in the vertical GRF values and the propulsive effort calculated from the anterior-posterior GRF were reported. Additionally, the shank muscles, Gastrocnemius, MG, and Soleus, SOL, activation level decreased in the presence of the applied force vector. These results show that the subjects reduced the trailing leg's effort during the push-off phase. Thus, our hypothesis that applied force at pelvis can help reduce the user's walking effort is accepted.

There is a great emphasis in the community to develop assistive devices for restoring the walking performances of individuals with lower limb impairments, such as lower limb amputees and elderly people. Passive and active mechanisms have been employed to intervene the human walking by exerting external forces at the lower limbs [5]–[15]. Currently, the external force profiles are decided based on simple mathematical models, such as inverted pendulum walking model, and gait kinetic patterns of healthy people [2]–[4]. It has been predicted that an impulsive push along the trailing leg immediately before the contra-lateral leg's HS can result in metabolically efficient walking pattern. Despite various efforts, a general consensus on the actuation timing and duration of the external forces that can reduce a user's walking effort has not been reached [12], [13]. In this context, our proposed methodology can be used to scientifically study the role of external forces in altering the walking effort. Using the proposed methodology external forces can be easily synchronized with a user's gait and can be applied at desired points during a gait cycle and for desired durations. We believe that these studies can develop better understanding to improve the current design and control of lower limb exoskeletons and prosthesis.

The results of the current experiment showed that the metabolic cost of walking did not reduce even though the subjects applied less effort from the trailing leg during the stance phase. It is noteworthy to mention that the current design objective of assistive devices is to reduce the overall metabolic cost of walking. Based on the current study, we suggest that the criteria of evaluating a device primarily on the metabolic cost could be misleading. Notably, it was reported in [13] that a powered ankle push-off prosthetic device which reduces the metabolic cost of walking does so by reducing the leg's effort during the swing phase and not by reducing the overall COM work. This was because the stance phase muscle activity comprises a large portion of the overall walking effort as opposed to the swing phase. These results indicate that the prediction of simple walking models may not necessarily translate to

a situation where external interventions are applied. One potential reason for this is that humans adapt their walking patterns under different external conditions. The addition of orthotic or prosthetic systems to apply forces at lower limbs modify the human walking dynamics, in particular due to the added mass on the leg and mismatch in timing when the external forces are applied [6], [12], [13].

An experimental methodology which can be used to test the effects of a force profile on a user's walking effort may be beneficial during the design process. In the current study, external forces were applied to assist the trailing leg during the step to step transition. This could be applicable to individuals with weaker ankle joints who demonstrate abnormal push-off forces from the trailing leg. In a similar way, the current methodology can be adapted to develop force interventions for an individual needing assistance while braking the COM with the leading leg or for an individual with weaker hip joint. However, at present, we have not tested the proposed force profile with a lower limb assistive device. We propose to conduct these studies in future.

V. Conclusion

In this work, we developed a novel experimental methodology to apply gait synchronized external forces at the pelvis using a cable-driven robot. An adaptive frequency oscillator scheme was used to plan the cable tension values to adapt the timing of force actuation to the gait frequency during walking. The results of the human experiment showed that external forces applied at the pelvis can be programmed to assist a user's trailing leg during the push-off phase of walking. We believe that studies conducted with the presented setup and control methodology can provide better understanding in designing assistive devices for individuals who spend higher than normal effort during walking.

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Fig. 1.

(a) The cable configuration used during the A-TPAD experiment is shown. Four cables were used to apply the desired sagittal plane forces. The hip belt was connected to the frame top by two cables and to the frame bottom by other two cables. (b) A healthy individual's pelvic vertical motion and acceleration profile over a gait cycle during steady-state walking. (c) The desired sagittal plane external force pattern over a gait cycle that was used during the study. The F_Y and F_Z values were comprised of varying terms (F_{YV} and F_{ZV}) that appeared briefly during the gait cycle and constant terms (F_{YC} and F_{ZC}) that were necessary to maintain positive cable tension during the experiment.

Fig. 2.

(a) The vertical pelvic acceleration profile from the accelerometer (θ_d) and the estimated acceleration values ($\hat{\theta}$) are shown for a section of the experiment. (b) The foot switches data from the heel and toe sensors of both legs is shown. The left and right gait cycle phase values (ϕ_{RGC} and ϕ_{LGC}) as calculated using Algorithm 2 are also presented for a section of the experiment

Fig. 3.

(a) Weight normalized ground reaction force components in the anterior-posterior (GRF_Y) and vertical (GRF_Z) directions. (b) The time integral of positive GRF_Y values (I_B , braking impulse), time integral of negative GRF_Y values (I_P , propulsive impulse), peak GRF_Z values, and time integral of GRF_Z values (I_{FN}) during a gait cycle are presented for the group. Significant changes were observed in the right leg's I_P , I_B , peak GRF_Z , and I_{FN} values between trials BL-T1 and BL-T4.

Fig. 4. Weight normalized volumetric oxygen consumption rate (VO_2/kg) for each subject over the ten minutes of the baseline and training sessions.

Fig. 5.

(a) Lower limb EMG envelopes during different experimental trials for a representative subject. (b) EMG peak and RMS values during the major burst over different trials of the experiment. Lower SOL and MG values were reported during the training trials.

TABLE I

Normalized volumetric oxygen consumption rate (VO2/kg).

	Avg. VO2/kg						Avg. HR	
	Full trial		1 st min		10 th min		Full trial	
	BL	T	BL	T	BL	T	BL	T
Sub 1	11.5	10.1	11.8	10.5	11.7	9.9	83.2	83.1
Sub 2	10.1	9.8	10.4	11.0	10.0	9.5	88.5	87.6
Sub 3	11.0	11.9	12.2	12.1	11.8	11.2	101.4	103.6
Sub 4	11.9	12.8	12.4	13.4	12.4	13.6	111.4	111.2
Sub 5	9.9	9.2	10.4	9.5	9.9	9.3	91.5	86.3
Avg.	10.9	10.7	11.5	11.3	11.2	10.7	95.2	94.5