A Novel Design of a Cable-driven Active Leg Exoskeleton (C-ALEX) and Gait Training with Human Subjects

Xin Jin

Submitted in partial fulfillment of the
requirements for the degree of
Doctor of Philosophy
in the Graduate School of Arts and Sciences

COLUMBIA UNIVERSITY

2018

ABSTRACT

A Novel Design of a Cable-driven Active Leg Exoskeleton (C-ALEX) and Gait Training with Human Subjects

Xin Jin

Exoskeletons for gait training commonly use a rigid-linked "skeleton" which makes them heavy and bulky. Cable-driven exoskeletons eliminate the rigid-linked skeleton structure, therefore creating a lighter and more transparent design. Current cable-driven leg exoskeletons are limited to gait assistance use. This thesis presented the Cable-driven Active Leg Exoskeleton (C-ALEX) designed for gait retraining and rehabilitation. Benefited from the cable-driven design, C-ALEX has minimal weight and inertia (4.7 kg) and allows all the degrees-of-freedom (DoF) of the leg of the user. C-ALEX uses an assist-as-needed (AAN) controller to train the user to walk in a new gait pattern.

A preliminary design of C-ALEX was first presented, and an experiment was done with this preliminary design to study the effectiveness of the AAN controller. The result on six healthy subjects showed that the subjects were able to follow a new gait pattern significantly more accurately with the help of the AAN controller. After this experiment, C-ALEX was redesigned to improve its functionality. The improved design of C-ALEX is lighter, has more DoFs and larger range-of-motion. The controller of the improved design improved the continuity of the generated cable tensions and added the function to estimate the phase of the gait of the user in real-time.

With the improved design of C-ALEX, an experiment was performed to study the effect of the weight and inertia of an exoskeleton on the gait of the user. C-ALEX was used

to simulate exoskeletons with different levels of weight and inertia by adding extra mass and change the weight compensation level. The result on ten subjects showed that adding extra mass increased step length and reduced knee flexion. Compensating the weight of the mass partially restored the knee flexion but not the step length, implying that the inertia of the mass is responsible for the change. This study showed the distinctive effect of weight and inertia on gait and demonstrated the benefit of a lightweight exoskeleton.

C-ALEX was designed for gait training and rehabilitation, and its training effectiveness was studied in nine healthy subjects and a stroke patient. The healthy subjects trained with C-ALEX to walk in a new gait pattern with 30% increase in step height for 40 min. After the training, the subjects were able to closely repeat the trained gait pattern without C-ALEX, and the step height of the subjects increased significantly. A stroke patient also tested C-ALEX for 40 minutes and showed short-term improvements in step length, step height, and knee flexion after training. The result showed the effectiveness of C-ALEX in gait training and its potential to be used in stroke rehabilitation.

Contents

List of Figures

List of Tables

Acknowledgements

The path to a doctorate is a long journey, and I certainly could not reach the finishing line without all the support I received from my professors, colleagues, and my family. I would like to take this opportunity to express my gratitude to all of them.

First and foremost, I would like to express my most sincere gratitude to my adviser and mentor, Dr. Sunil Agrawal, for his continuous support of my research, for his patience, motivation, enthusiasm, and immense knowledge. From him, I learned how to do good research, and how to be a good engineer.

Secondly, I would like to appreciate all the insightful comments and advice from my dissertation committee members: Dr. Matei Ciocarlie, Dr. Joel Stein, Dr. Andrew Gordon and Dr. Gerard Ateshian. I am also very grateful to all the professors who have taught me during the years of my study. The knowledge I learned was directly put to use in this thesis.

Thirdly, I need to thank all the current and past members of the ROAR Lab at Columbia University. I am very fortunate to have the chance to know so many smart people, to work with them and to learn from them. In particular, I want to appreciate Dr. Ying Mao, Antonio Prado, Yusheng Cai, Rand Hidayah and Lauri Bishop who spent a significant amount of their time and worked closely with me on this project. Their effort pushed this

project to go much further than I could have done by myself.

Last but not least, I would like to thank my parents for their unconditional support to me throughout my life, and my wife Xinyue Wang, for her encouragements, companionship, and love.

Introduction

Exoskeletons in Rehabilitation

Exoskeletons are wearable robotic devices designed to provide assistance or support to the human users during functional movements. Various exoskeletons have been designed for different purposes with varying levels of success. Some exoskeletons were designed to augment human capability [1], [2], some were designed to assist people with disability [3], [4], while some exoskeletons were designed for training and rehabilitation [5], [6].

Many of the exoskeletons designed for physical therapy training specifically targeted at stroke rehabilitation [7]–[9]. There are approximated 795,000 cases of stroke occurring in the United States each year, and stroke is the among leading causes of long-term disability in the United States [10], [11]. The stroke survivors with disabilities require physical therapy to regain the motor functionalities lost due to stroke. Traditional physical therapy usually involves physical therapist manually moving the body of the stroke patient, which is labor intensive. Compared with tradition physical therapy, using robotics such as exoskeletons to assist physical training can save the therapists from the tedious manual labor, and gives the patients longer, more intensive, and more consistent training [12].

Various leg exoskeletons have been designed and tested for gait rehabilitation of

stroke patient. These exoskeletons can be roughly categorized as stationary systems and mobile systems. The stationary systems generally aim to assist and automate the body weight supported treadmill training, during which the stroke patient walks on a treadmill with body weight support with the exoskeleton assisting the movement of the legs of the patient. Lokomat [1] is a commercially available stationary exoskeleton with integrated treadmill, and has been widely used in rehabilitation facilities. It is a bilateral leg exoskeleton that can move the leg of the user in a pre-programmed gait-like pattern [13]. Studies have shown that training with Lokomat helps to improve gait functions in stroke patients [14], but there are debates on whether it is better or worse compared with traditional physical therapy performed without the exoskeleton [14]–[16]. Some exoskeletons have been developed for research purposes and are not yet commercially available, such as ALEX and its later version ALEX II and III [17]–[19]. All of these are exoskeletons for treadmill based walking. They have been tested with stroke patients and showed positive results [6], [20].

Mobile leg exoskeletons include Ekso [2], ReWalk [3], HAL [4], Vanderbilt exoskeleton [21] and H2 [22]. These exoskeletons are self-contained devices and allow the patients to ambulate over-ground, usually with the help of crutches or a walker to maintain balance. Most of these mobile exoskeletons were originally designed for walking assist on paraplegic users such as SCI patients. In terms of rehabilitation of stroke patients, Ekso currently is undergoing clinical trials on stroke patients [23]. HAL has been tested on stroke

[1]Hocoma, Switzerland

[2]Ekso Bionics, CA, USA

[3]Argo Medical Technologies, Israel

[4]Cyberdyne, Japan

patients in several studies [24]–[26], with positive training outcome.

Early exoskeletons used position control to move the leg of the patient [27]. This kind of control can provide all the necessary forces for the patients to complete a desired motion, therefore allows patients with severe impairments to move again. However, position control also allows patients to slack off during the training and simply rely on the exoskeleton to complete the motion [15], [28]. Active patient participation is critical in rehabilitation [29], therefore later exoskeletons used various strategies of force control that require patients to attempt the motion actively. The controller monitors the performance of the user and only assists the user when the performance is not adequate [17], [30]–[34]. These control strategies are referred to as assist-as-needed (AAN) controller. The AAN controllers commonly have a passive or transparent mode in which no assistance is provided. Under this mode, the exoskeleton simply follow the motion of the user but should not interfere in any way. The exoskeleton serves as a measuring device to evaluation the performance of the user under this mode. Studies have shown that AAN controller can improve training outcomes compared with position based control strategy [35].

Limitations of the Rigid-link Based Designs

Exoskeletons, by their name, usually include a mechanical "skeleton" that is made of linkages and joints. This very "skeleton" structure has two issues that may hinder the transparency of the exoskeleton. One issue is the weight and inertia of the mechanical structure. If not properly compensated by the exoskeleton, moving the weight and inertia

of the exoskeleton will burden the user, causing changes in the gait kinematics [36] and increasing the energy expenditure during gait [1]. Another issue with the mechanical "skeleton" is that it usually cannot accommodate all the degrees-of-freedom (DoF) of the human leg [6], [18], thus constraining the motion of the user. Both of these issues have been studied in the past with different methods proposed by the researchers to alleviate them.

The human leg is roughly 15% of the body weight [37]. The weight of a leg exoskeleton is typically comparable to the weight of the leg or even larger [38]. The weight and inertia of the exoskeleton will impede the natural motion of the user if left uncompensated [36]. Compensating the weight of the exoskeleton is easier because it only requires knowing on the pose of the exoskeleton which can be measured with ease. Weight compensation can be done either passively using springs [39] or actively through actuators [19]. Inertia compensation is harder because it requires the knowledge of the velocity and acceleration of the exoskeleton in real-time, which are usually not readily available. One solution to reduce the impact of inertia on the human user is to directly measure the interaction force between the human and the exoskeleton, and then compensate for it through feedback control [40]. However, non-collocated sensing and actuation may reduce the stability of the system [41], and this approach also requires expensive multi-axis force sensors at the human-exoskeleton interface. Another solution is to exploit the cyclic nature of the gait and estimate the speed and acceleration using adaptive frequency oscillator [42]. Their results showed improvement in transparency. However, a possible limitation of this approach is that the gait of an impaired patient may not be as repetitive as the gait of healthy subjects, and therefore the performance of the adaptive frequency oscillator

might be reduced.

Choosing the proper DoF is another challenge in the exoskeleton design. During walking, humans have motion in 6 DoFs at the pelvis, 3 DoFs at the hip joint and 1 DoF motion at the knee joint. Few exoskeletons can accommodate all these DoFs in the design, especially the DoFs at the pelvis. Limiting the DoF of pelvis changes the gait characteristics of the user [43], [44], alters the muscle activation pattern [45] and affects the gait-training outcome [39]. However, adding more DoFs to the exoskeleton is not a simple task. Adding DoF not only requires additional joints to allow the motion, but also requires additional mechanisms and actuators to compensate for the weight and inertia acting on the additional joint, which further increases the weight and inertia of the whole system and complicates the design. Several attempts have been made to allow more DoF at the pelvis. ALEX II has 5 DoFs in the pelvic motion, but all these DoFs are passive. PAM and POGO included 5 active and 1 passive DoF at the pelvis by the use of parallel mechanisms [33]. Lopes II included 3 active and 3 passive DoFs at the pelvis [46], and was able to place the actuators away from the moving parts to reduce inertia. All these three exoskeletons turned out to be large and bulky. Mobile exoskeletons, such as Ekso or ReWalk, has the advantage on the pelvic DoF. Their design inherently allows all the pelvic motion.

Cable-driven Designs

Despite the drawbacks of the mechanical "skeleton," it serves the important purpose of transferring power from the actuators to the user. Recently, an emerging approach in ex-

oskeleton design seeks to use cables to transfer actuation to the user [47]–[52]. The rigid mechanical "skeleton" is replaced by several lightweight cables anchored to the body of the user. Some of these cable-driven exoskeletons are made out of soft fabric and are sometimes referred to as "exosuits" [51], [52]. The cable-driven design can be made lightweight and compact with no restrictions on joint movement, and are therefore ideal for highly dynamic motions such as gait assistance.

Walsh *et al.* at Harvard University have developed an exosuit with cable and fabrics [52]. The exosuit includes a single cable at the back of each ankle joint to provide plantar flexion torques during walking [53], and some variations of its design also include a cable at the back of each hip joint to provide hip extension torque [54]. This exosuit generates prescribed tension profile in the cable based on gait events [55], [56]. The exosuit is designed for gait assistance and reducing metabolic cost of the user in loaded and unloaded walking. It has also been tested on stroke patients and showed improvements in gait symmetry [57]. Collins *et al.* at Carnegie Mellon University developed several cable-driven ankle exoskeletons aiming at reducing metabolic cost during walking. There is a passive design that used a clutch and spring to store energy during the stance phase of the gait and release during ankle push off [58], and an active design that used human-in-the-loop tuning to optimize the timing and magnitude of the assistance delivery [48].

Presently, exoskeletons for gait training are still based on the rigid-link designs and suffers from the issues of heavy weight, large inertia, and restricted joint motion [7], [59], [60]. Cable-driven approach is a good, though unproven, alternative approach for training exoskeleton designs. However, to be used in gait training, several improvements on the cable-driven design should be made: current exosuits designs usually target a single

joint and can only provide unidirectional torque, while training exoskeletons usually have bidirectional torque generation at multiple joints; current exosuits generate assistance based on the gait timing, while training exoskeletons commonly use an assist-as-needed controller as mentioned earlier.

To bring the benefits of cable-driven designs into gait training exoskeletons, a Cable-driven Active Leg Exoskeleton, named C-ALEX is designed and tested. C-ALEX is a unilateral leg exoskeleton designed for gait training, eventually targeting at stroke patients. Like the other exosuit designs, C-ALEX does not have a mechanical rigid-link "skeleton." It uses cables for actuation, and the cables are anchored to the leg of the user using lightweight, 3D printed cuffs. Since C-ALEX has no mechanical joints, it does not restrict any DoF of the human user. Using four cables, C-ALEX can provide bidirectional flexion-extension torque assistance at the hip and knee joints. C-ALEX uses an assist-as-needed controller to help the user adapt to a new gait pattern.

Overiew of the Thesis

The goal of this thesis is to demonstrate the design of C-ALEX and present its experiment results on healthy subjects and a stroke patient. The rest of this thesis are arranged as follows: Chapter 1 introduces an preliminary design of C-ALEX and a study that verifies its effectiveness in changing the gait of the user. Chapter 2 shows a redesign of C-ALEX that improves its various functionality. This improved C-ALEX was used in the subsequent studies. Chapter 3 presents a study that investigates the effect of exoskeleton weight and inertia on the natural gait of the user. Chapter 4 demonstrates the effectiveness of

C-ALEX in gait training of healthy subjects. Chapter 5 evaluates C-ALEX on a stroke patient to test the feasibility of using C-ALEX in stroke rehabilitation. The last Chapter concludes the thesis and discusses possible future directions.

Chapter 1

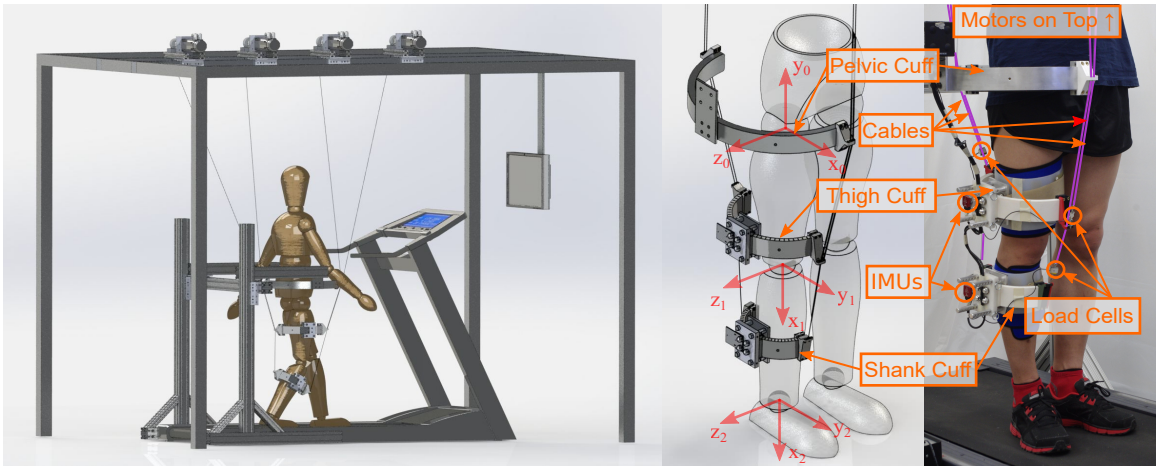
A Preliminary Design of C-ALEX and Evaluation on Healthy Subjects

This chapter presents the preliminary design of the cable-driven active leg exoskeleton (C-ALEX). Despite being a preliminary design, this design already incorporates the main feature of a cable-driven exoskeleton: a lightweight and restriction-free structure without any mechanical links or joints. The controller of the preliminary design for gait assistance is also presented, and an experiment with healthy subjects was conducted to evaluate the effectiveness of the preliminary design in assisting users to follow a new gait pattern. The experiment demonstrates the feasibility of the cable-driven design for lower extremity usage, and also points the directions for future improvements of C-ALEX.

1.1 The Mechanical Design

Exoskeleton Design

Fig. 1.1 shows the preliminary design of C-ALEX. The preliminary design of C-ALEX includes an overhead frame to support the motors, a treadmill for the user to walk on, and exoskeleton attached to the leg of the user. The exoskeleton mainly consists of three cuffs: the pelvic cuff, the thigh cuff and the shank cuff. No rigid link or joint is used



(A) (B) (C)

Figure 1.1: The Preliminary Design of C-ALEX. (A) An overview of the preliminary design. (B) A closer view of the cuffs of the preliminary design with a local coordinate system for each segment of the leg. (C) C-ALEX attached to the leg of a user.

within the exoskeleton. The pelvic cuff is fixed to a height-adjustable external support frame. The thigh and the shank cuffs are tightly connected to the user's thigh and shank, respectively. To create a secure connection between the thigh and shank cuffs and the leg, a layer of elastic band is first placed onto the user's leg. An orthotics with velcro liners is then strapped on top of the elastic band. The cuffs are attached at the lateral side of the orthotics. The lateral distance between the cuff and the leg can be adjusted. To reduce the weight of the exoskeleton on subject's leg, the thigh and shank cuffs are primarily made of 3D printed ABS plastic with sparse interior. The overall weight of the thigh and shank cuffs are 0.60kg and 0.54kg, respectively.

Four cables are used to actuate the exoskeleton. The cables are pre-stretched Nylon coated steel wires. All four cables are routed through the pelvic cuff. Two of these are attached to the thigh cuff, and the other two are routed through the thigh cuff and attached to the shank cuff. These four cables actuate two degrees-of-freedom of the user's

leg: the hip flexion/extension and the knee flexion/extension. The DH parameters of the exoskeleton are shown in Table 1.1. In the DH parameters, q_1 is the hip flexion angle and q_2 is the knee flexion angle. L_{th} stands for thigh length, and L_{sh} stands for shank length. These two parameters vary between individuals, and therefore need to be measured and provided for each subject. The kinematics of C-ALEX are derived through the DH parameters and homogeneous transformations. All the cable routing points on the cuff have Teflon liners to reduce the friction between the cable and the cuff. The cable routing points are designed to be able to slide along the cuff to change the cable routing. Additional cable routing points can be added to accommodate extra cables to increase the controlled degrees-of-freedom.

Table 1.1: DH Parameters of the Preliminary Design

Link	a	d	α	θ
1	L_{th}	0	0	$q_1 - \pi/2$
2	L_{sh}	0	0	$-q_2$

L_{th}: thigh length; L_{sh}: shank length.

Actuation and Control System

Each cable in the exoskeleton is driven by a AKM43 servo motor [1] with custom made cable reels. The motors are supported by the overhead frame. The cable reels are used to help proper winding of the cables. The motors are powered by AKD servo drives [1] operating in torque mode. A LSB200 load cell [2] is connected to the end of each cable to measure the tension in the cable, with CSG110 signal conditioners [2] to amplify the load

[1]Kollmorgen, VA, US

[2]Futek, CA, US

cell signal. There is a VN-100 inertia measurement unit (IMU)[1] mounted on the thigh cuff and shank cuff to measure the hip and knee angle of the user during walking. A PXIe-8135 controller[2] is used for real-time control and data acquisition of the exoskeleton. The controlling software is developed in LabVIEW[2].

1.2 The Controller for Gait Assistance

The Dynamical Model of C-ALEX

The dynamic equations of motion of C-ALEX are derived through the Lagrangian method in Mathematica [3]. As mentioned in Section 1.1, C-ALEX is modeled as a two-link serial chain in the sagittal plane of the leg as shown in Fig. 1.1. Based on this kinematic model, the dynamical equation of C-ALEX can be written in the following form:

$$D(q)\ddot{q} + C(q, \dot{q})\dot{q} + G(q) = \tau_c - \tau_h \,, \tag{1.1}$$

where $q = [q_1, q_2]^T$ is the vector of joint angles with q_1 and q_2 represent the angles of hip flexion and knee flexion, respectively; $D(q)$ is the inertia matrix; $C(q, \dot{q})$ is the vector of Coriolis and centripetal terms; $G(q)$ is the vector of gravity terms; τ_c is the actuation torques generated by the cables at each joint, and τ_h is the torques that C-ALEX applies to the human user. The D, C, and G terms on the left hand side of the equation are obtained from the CAD model of C-ALEX and the measurements of the leg of the user.

[1]VectorNav, TX, US

[2]National Instrument, TX, US

[3]Wolfram, IL, US

Equation (1.1) can be rewritten as:

$$\tau_h = \tau_c - (D(q)\ddot{q} + C(q,\dot{q})\dot{q} + G(q)) \,, \tag{1.2}$$

which shows that the torque applied to the human τ_h equals the actuation torque τ_c minus the gravity G and dynamics D and C of the exoskeleton. Ideally, the user of the exoskeleton should only receive the intended assistance without any disturbance due to the weight and inertia of the exoskeleton, *i.e.* :

$$\tau_h = \tau_a \,, \tag{1.3}$$

where τ_a is the intended assistive torque to train the human user. As a result:

$$\tau_c = \tau_a + D(q)\ddot{q} + C(q,\dot{q})\dot{q} + G(q) \,, \tag{1.4}$$

which shows that the exoskeleton should be able to compensate for its dynamics ($D(q)\ddot{q}+C(q,\dot{q})\dot{q}$) and gravity ($G(q)$) in addition to providing assistance in order to be fully transparent. The gravity of the exoskeleton can be compensated easily as it only requires the knowledge of the joint angles q. The dynamics of the exoskeleton are harder to compensate for as they involve joint velocity $\dot{q}$ and joint acceleration $\ddot{q}$, neither of which are directly measured by C-ALEX. Walking is a highly dynamical motion, therefore $\dot{q}$ and $\ddot{q}$ cannot be simply ignored. If an exoskeleton has a large inertia, it will require a large amount of torques to follow the motion of the leg. However, given the light-weight nature of C-ALEX, it does not require significant torques to move the cuffs on the leg, and there-

fore the dynamical terms in Eq. (1.2) were neglected. Then the torque τ_c to be generated by cables equals to:

$$\tau_c = \tau_a + G(q)\,. \tag{1.5}$$

C-ALEX uses an "Assist-As-Needed" control strategy to train the user to follow a prescribed ankle path. The exact method to generate the assistive torque τ_a is detailed in Section 1.2, with some improvements introduced in Section 2.2. C-ALEX has a "transparent" mode, in which case the exoskeleton does not provide any assistance to the user, *i.e.* the assistive torque $\tau_a = 0$. Under this condition, the actuation torque τ_c only compensates for the gravity of the exoskeleton:

$$\tau_c = G(q)\,. \tag{1.6}$$

Cable Tension Planning

In C-ALEX, the actuation torque τ_c is generated by the four cables. The relationship between cable tension T and actuation torque τ_c can be found by:

$$J_C(q)T = \tau_c\,, \tag{1.7}$$

where $T = [T_1, T_2, T_3, T_4]^T$ represents the tensions in each cable. $J_C(q)$ is the Jacobian matrix relation cable tension to joint torques. $J_C(q)$ can be found by the virtual work principle as explained in the following paragraph.

Fig. 1.2 shows the schematic of a two-link system actuated by a single cable. This

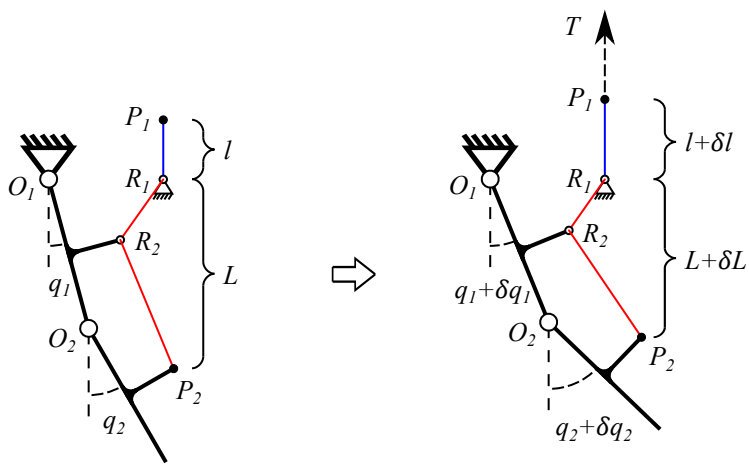


Figure 1.2: A Schematic of a Two-link System Actuated by a Single Cable. One end of the cable is free (P_1), while the other end of the cable is routed through point R_1 and R_2, and fixed to the two-link system at point P_2. q_1 and q_2 represent the joint angles of the two-link system; l denotes the length of the cable outside the two-link system (from P_1 to R_1); L denotes the length of the cable routed inside the two-link system (from R_1 to P_2). Force T pulls the cable at the free end (P_1) and results in changes in q_1, q_2, l and L.

schematic represents a simplified model of the cable-driven exoskeleton. One end of the cable is free while the other end of the cable is routed through a few routing points on the two-link system and attached to the distal link. Initially, the joint angles of the two-link system are q_1 and q_2, the length of the cable that is routed inside the two-link system is L, and the length of the cable remains outside the two-link system is l. A force T is applied to the free end of the cable and causing a virtual displacement of δl. As a result, the configuration of the two-link system changes. The angles of the two joints change by δq_1 and δq_2, and the length of the cable routed inside the two-link system changed by δL. Given that the total length of the cable ($l + L$) is constant, it is apparent that:

$$\delta L = -\delta l \,. \tag{1.8}$$

According to the principle of virtual work, the virtual work done by the force T on

the cable equals to the virtual work done by the generalized forces on the joints:

$$\delta W = T \cdot \delta l = \tau \cdot \delta q \,, \tag{1.9}$$

in which $\tau = [\tau_1, \tau_2]^T$ represents the generalized force, *i.e.* torques applied to the joints $q = [q_1, q_2]^T$. We also have:

$$\delta l = -\delta L = -\frac{\partial L}{\partial q}\,\delta q \,. \tag{1.10}$$

Substituting Eq. (1.10) into Eq. (1.9) gives:

$$\tau = -(\frac{\partial L}{\partial q})^T\, T = J_C(q)^T\, T \,. \tag{1.11}$$

Equation (1.11) shows that the Jacobian matrix relating the cable tensions to the joint torques can be found by the partial derivatives between the cable lengths and the joint angles. With four cables used in C-ALEX, the Jacobian matrix $J_C(q)$ in Eq. (1.7) is

$$J_C(q) = -\frac{\partial(L_1, L_2, L_3, L_4)}{\partial(q_1, q_2)} \,, \tag{1.12}$$

where L_i is the length of cable i measured from the routing point on the pelvic cuff to the final attachment point of the cable on the thigh or shank cuffs.

Equation (1.11) is of particular importance to the system as it provides the foundation to solve for the required cable tensions when some specific torques are desired at the joints. When torque τ_c is required at the joints, cable tensions T can be found by solving

the set of linear equations:

$$J_C(q)^T\, T = \tau_c\,. \tag{1.13}$$

Equation (1.13) is underdetermined as the number of cables is larger than the degrees-of-freedom of the system.

The range of tensions in the cables is limited. Since cables can only pull but not push, it is impossible for the tensions in the cables to be negative. In the actual system, due to the existence of friction along the cables, the minimum tension in a cable needs to be set above a positive value to keep the cable taut. Also, because the motors connected to the cables can only produce limited torque, there is a maximum limit on cable tensions as well. Therefore, T should satisfy:

$$T \in [T_{min}, T_{max}]\,. \tag{1.14}$$

In the case of C-ALEX, the cable tension range is set in between 7N and 70N.

Using Eq. (1.13) and Eq. (1.14) as constraints, an optimization problem can be formulated to find a proper set of cable tensions to generate desired torques. We use a quadratic objective function: $f(T) = T^T\, T$ for the optimization problem, which minimizes the norm of cable tension vector. The advantage of using quadratic programming over linear programming is that the solution to T will change more continuously when the Jacobian matrix $J_C(q)$ in the equality constraint Eq. (1.13) changes. This will help to avoid abrupt cable tension changes when the leg moves from one configuration to another.

Overall, the cable tension planning problem can be formulated as a quadratic pro-

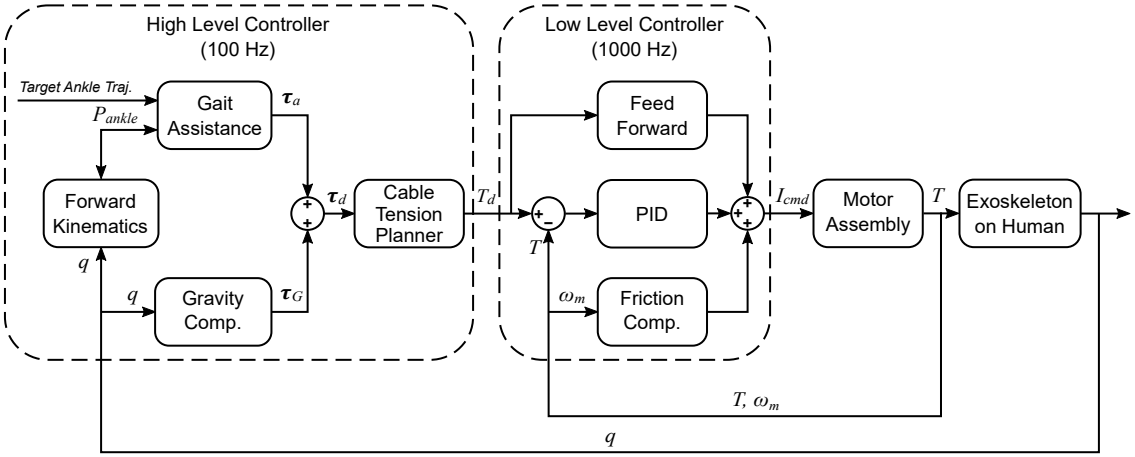


Figure 1.3: A Block Diagram of the Controller of C-ALEX.

gramming problem:

$$\min f(T) = T^T T$$

$$\text{s.t.} \quad J_C(q)^T \cdot T = U \quad \text{and} \quad T \in [T_{min}, T_{max}] \,.$$

LabVIEW provides a quadratic programming solver that uses the active set method. This solver is used to solve the above problem in real time. The disadvantage of quadratic programming is that it is time consuming to solve. However, our controller is capable of solving the above problem at 100Hz without any delay.

The "Assist-as-needed" Controller

This exoskeleton is aimed at rehabilitation and therefore the controller uses an "assist-as-needed" strategy. The goal of the controller is to assist the ankle point of the user of the exoskeleton to move on a prescribed target path. The controller creates a tunnel-like force-field around the target path. If the end effector (ankle point) deviates from the target

path, the controller acts as a spring and pulls the end effector back to the target path. A two-level controller is implemented as shown in Fig. 1.3: A high level force-field controller that uses the position feedback of the exoskeleton to dictate the necessary cable tensions to create the force field, and a low level cable tension controller that controls the motors to produce the desired cable tensions using feedback from the load cells on the cables.

The high level force-field controller generates a force F at the ankle point that has two components:

$$F = F_n + F_t. \tag{1.15}$$

F_n is normal to the target path, and it will push the ankle point closer to the path. F_t is tangential to the target path, pointing to the direction of movement, and it provides a gentle push to move the ankle point along the target path. The magnitude of F_n, F_t is calculated as:

$$\begin{aligned} \|F_n\| &= K_n \cdot (1 - e^{-(\frac{2d}{r_n})^2}) \\ \|F_t\| &= K_t \cdot e^{-(\frac{2d}{r_t})^2}, \end{aligned} \tag{1.16}$$

in which K_n and K_t are the gains of the force field, d is the distance from the ankle point to its nearest point on the target path. Equation (1.16) effectively creates two tunnels around the target path with diameters of r_n and r_t respectively. For the normal force F_n, the magnitude roughly equals to K_n outside the tunnel and gradually deceases to 0 N inside the tunnel. For the tangential force F_t, the magnitude is 0 N outside the tunnel and gradually increases to K_t inside the tunnel. Fig. 1.4 shows the change of F_n and F_t as a function of the normal distance from the target path d, and a qualitative sketch of the force field around a target path.

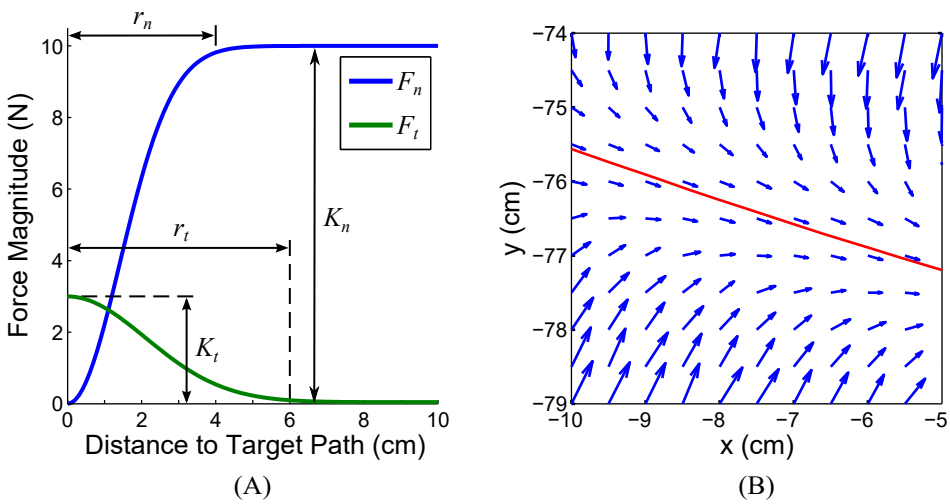


Figure 1.4: Force Field Controller of C-ALEX. (A) The magnitude of the force field as a function of the normal distance to the target path. (B) A portion of the force field around a target path. The red line is the target path, and the blue arrows show the directions and magnitudes of the force field.

The force-field controller has two modes: the transparent mode and the assistive mode. The transparent mode is used for measuring the natural gait of the user. In the transparent mode, C-ALEX will only balance its own weight and will not provide any assistance to the user. The assistive mode is used for assisting the user to track the target ankle path. In the assistive mode, C-ALEX generates the aforementioned force field besides compensating for its own weight. The required joint torque τ_c in the assistive mode can be found by:

$$\tau_c = J_e(q)^T F + G(q) \,, \tag{1.17}$$

where F is the force-field force in Eq. (1.16) and $J_e(q)$ is the Jacobian matrix of the end effector.

With the desired joint torque τ_c obtained, the force-field controller uses the cable tension planning explained in previous subsection to calculate the desired tension for each

cable and send it to the low level controller.

The low level controller includes three parts: A feed-forward part using the motor constant, a friction compensation part using the motor's friction-speed model, and a close loop PID controller using the feedback from the load cells in the cables. Together, the low level controller is able to control the motors to generate tensions that closely follow the desired tensions calculated from the high level controller.

1.3 Evaluation on Healthy Subjects

An experiment was conducted to evaluate the performance of C-ALEX. The goal of the experiment was to verify that C-ALEX with the force field controller can assist the user to track a target ankle path different from his or her natural path. In the experiment, the subjects tracked a target ankle path with C-ALEX in both the assistive mode and the transparent mode. The hypothesis was that the subjects would have better path tracking performance in the assistive mode than in the transparent mode.

The setup of the experiment is shown in Fig. 1.5. Six healthy subjects participated in the experiment. The subjects were all male, aged between 20 and 35 years. Fig. 1.6 shows the protocol of the experiment. At the beginning of each experiment, C-ALEX was fitted onto the right leg of the subject, and then the experimenter took measurements of the length of the thigh and the shank of the subject as well as the locations of the cuffs. These measurements were provided to the controller. Each experiment was divided into 4 sessions: the familiarization session, the baseline session, the assistive (AS) session and the transparent (TR) session. Each session was 4 minutes long, and a 2 minutes break was

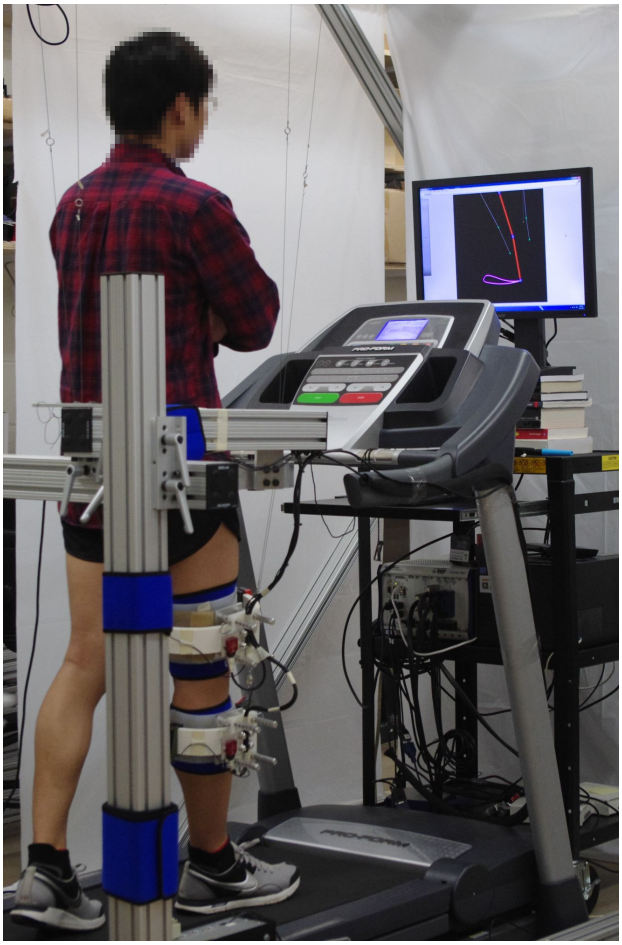


Figure 1.5: The Experimental Setup to Evaluate the Preliminary Design of C-ALEX. The subject was walking on a treadmill with C-ALEX attached to the leg. The screen placed in front of the subject was used to display the current pose of the leg and the target ankle path.

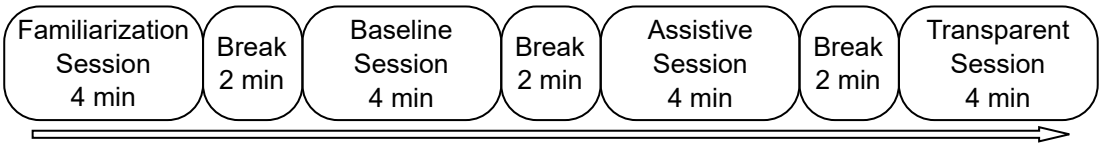


Figure 1.6: The Experiment Protocol to Evaluate the Preliminary Design of C-ALEX.

given between each session.

The experiment started with the familiarization session, during which C-ALEX was put into the transparent mode. The subject was instructed to walk on a treadmill to get familiar with walking with C-ALEX. The speed of the treadmill was adjusted to the com-

fortable walking speed of the subject. Following the familiarization session was the baseline session, during which C-ALEX stayed in the transparent mode. The subject was instructed to walk naturally during this session. The natural gait pattern collected during this session was later used to generate the target ankle path in the AS and the TR sessions. The screen in front of the subject was turned off during the first two sessions.

During the baseline session, the hip and knee joint angles and the ankle path were recorded. Data during the first and the last minutes were discarded, and the remaining data were cut into gait cycles at the anterior most point of the ankle path and averaged across the gait cycles to obtain the averaged joint angles in a single gait cycle. The averaged joint angles were then reduced by 20% to create an ankle path that is both shorter and shallower than the baseline path. The top figure of Fig. 1.7 shows the baseline ankle path (black) and the modified ankle path (red) from a representative subject. This modified ankle path was then used as the target ankle path in the AS and the TR sessions.

During the AS and the TR sessions, the screen in front of the subject was turned on to display the current configuration of the leg and the target ankle path, as shown in Fig. 1.5. The subject was instructed to try to walk as closely to the target path as possible during these two sessions. During the AS session, C-ALEX was put into the assistive mode. The force-field gains in Eq. (1.15) were set as $K_n = 20, r_n = 0.1, K_t = 3, r_t = 0.05$ and were consistent across all subjects. After AS session was the TR session, during which C-ALEX was put into the transparent mode. After the experiment, the joint angles and ankle path during each session were analyzed.

The ankle path during the AS and the TR sessions were recorded, cut and averaged in the same way as the ankle path during the baseline. The average ankle path during

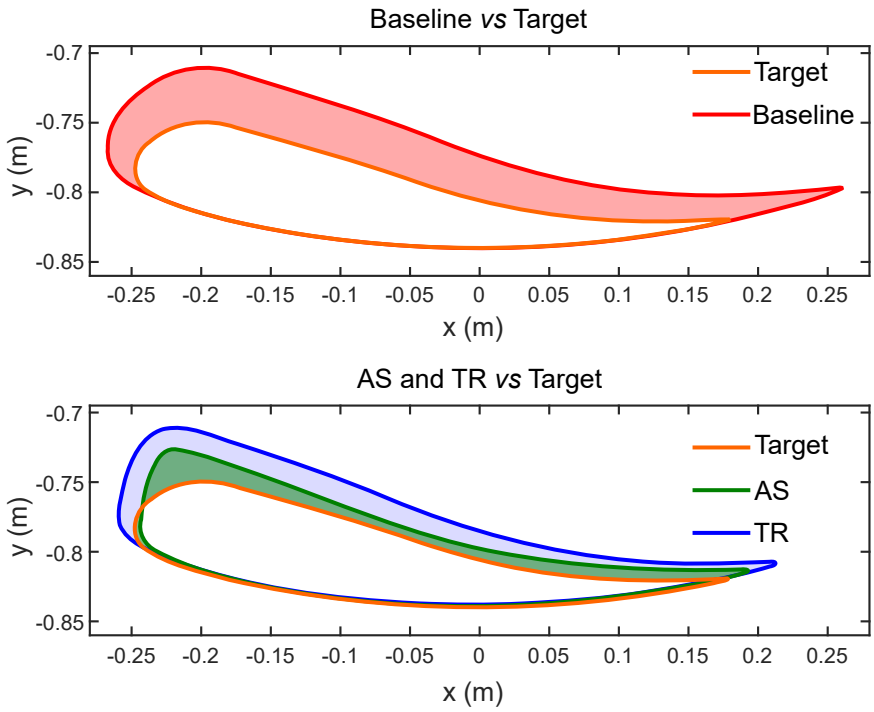


Figure 1.7: Ankle Path from a Representative Subject. The grey shaded area in the top figure is the deviation area between the baseline path and the target path. The blue and the green shaded areas in the bottom figure are the deviation area of the TR path and the deviation area of AS path, respectively.

the AS and the TR sessions of the same representative subject was plotted in the bottom figure of Fig. 1.7. It can be observed from the figure that the AS path is closer to the target path than the TR path, which demonstrates the effectiveness of the force-field controller of C-ALEX.

To quantify the effectiveness of C-ALEX, the normalized tracking error [61] of the AS path and the TR path was calculated and compared. Fig. 1.7 shows the deviation area of the baseline path (gray shaded area), the deviation area of the AS path (green shaded area) and the deviation area of the TR path (blue shaded area). The normalized tracking error of the AS (TR) path is calculated as the ratio between the deviation area of the AS (TR) path and the deviation area of the baseline path, i.e. the green (blue) shaded

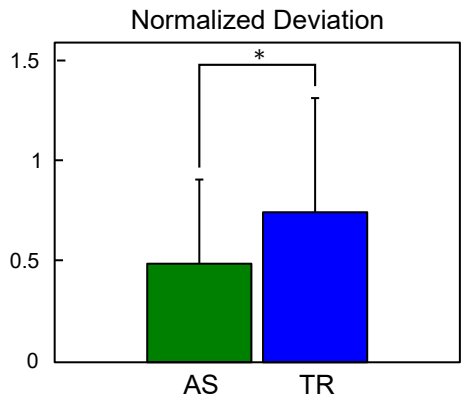


Figure 1.8: The Normalized Tracking Errors of the AS and TR session. The error bars represent standard deviation. "$*$" indicates significant difference at $p < 0.05$ level.

area divided by the grey shaded area. A smaller normalized tracking error suggests that the path closely overlaps the target path. Fig. 1.8 shows the normalized tracking error average across all subjects. The average normalized tracking error of the AS sessions is 0.493 ± 0.421 (mean±SD), and the average normalized tracking error of the TR sessions is 0.751 ± 0.572 (mean±SD). Wilcoxon signed rank test was used to compare the normalized tracking error in AS sessions and in TR sessions. The result shows that the normalized tracking error in AS sessions is significantly smaller than that in TR sessions ($p = 0.031$). This result validates our hypothesis that C-ALEX with force field can help the subject to follow a prescribed ankle path.

1.4 Conclusion

In conclusion, this chapter presented the preliminary design of the Cable-driven Active Leg EXoskeleton (C-ALEX) for human gait training. The exoskeleton is controlled by a force-field controller, which is able to assist the motion of the leg and help the ankle to

follow a desired path. An experiment with six subjects wearing C-ALEX was performed to validate the device and the controller. The results showed that the force-field controller is able to help the subject better track a prescribed ankle path.

Several limitations of the preliminary design of C-ALEX were identified: (i) the pelvic cuff of the exoskeleton is fixed to the ground, which restricts the pelvic motion of the user; (ii) the range of motion of the leg is limited by the cable routing; (iii) the current cable routing can only provide assistive force in the sagittal plane. These problems are rectified in the next design iteration of C-ALEX.

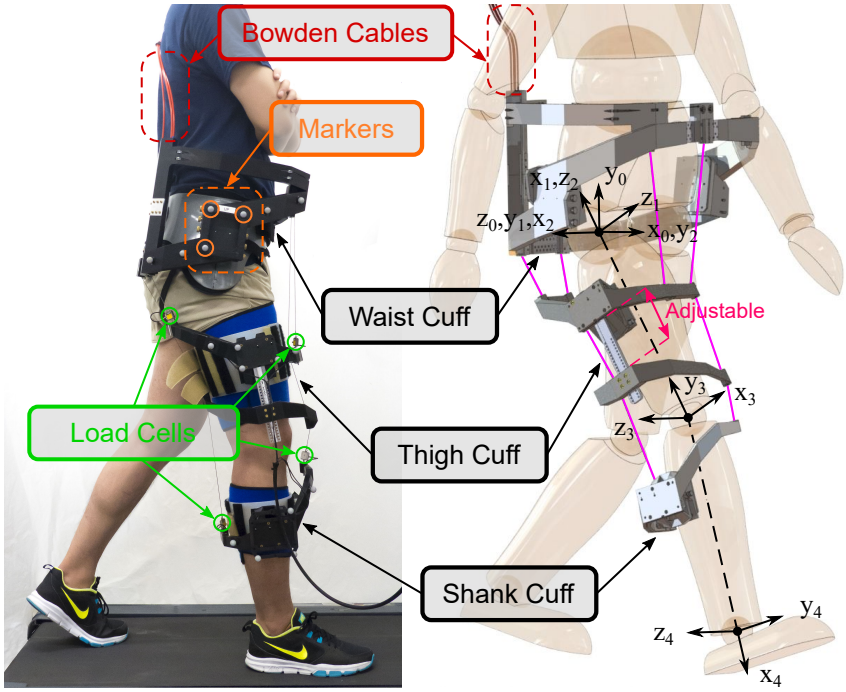
Chapter 2

An Improved Design of C-ALEX

After identifying the limitations of the preliminary design, an improved version of C-ALEX was designed and built. The cuffs of this new C-ALEX are completely redesigned to give the user more degrees-of-freedom and a greater range of motion. The controller has also several improvements to address the wrench feasible workspace limitation that is common in cable-driven robots.

2.1 Improvements in the Mechanical Design

The new design of C-ALEX shares some similar cable-driven characteristics as the previous preliminary design. Fig. 2.1 shows the CAD model of the new C-ALEX and the actual fabricated device worn by a user. The key feature of C-ALEX is its low weight. The cuffs of the new C-ALEX are redesigned to further reduce its weight. The cuffs are 3D printed with a sparse interior using ABS plastic, same as the preliminary design. The pelvic, thigh, and shank cuff weighs 2.7 kg, 1.0 kg and 0.6 kg, respectively. The pelvic cuff is largely stationary during walking, therefore the total moving weight of C-ALEX (thigh and shank cuff combined) is only 1.6 kg. The light-weight cuffs add minimal weight and inertia to the leg, thus create minimal disturbance to the gait of the wearer. Similar to the preliminary design, there are no mechanical links and joints to be attached to the human



(A) (B)

Figure 2.1: Improved Design of C-ALEX. (A) C-ALEX attached to the leg of a subject. Load cells at the end of each cable are circled in green. Orange circles show the reflective markers on the pelvic cuff for motion tracking. More of these markers are placed on the thigh and shank cuffs as well. (B) A CAD model of the exoskeleton with local coordinate system of the kinematic model.

leg in this new design of C-ALEX. The joint-free design allows the leg of the wearer to move freely in all its natural degrees-of-freedom without any constraints. Besides keeping the lightweight and joint-free feature, the new C-ALEX has two major improvements over the preliminary design: (i) a redesigned cable routing that gives larger and more consistent maximum range of motion (RoM), and (ii) a new pelvic cuff with Bowden cables that allows constraint-free pelvic movement.

The maximum RoM of C-ALEX is limited by the distance between the cables and the human limb. If the distance is too small, cables may come in contact with the human body during motion. If the distance is too large, the design becomes bulky and cumbersome.

In this new C-ALEX design, we redesigned the cuffs and strategically routed the cables to achieve a large RoM that is sufficient for gait training while keeping a compact cuff profile. Besides being larger, the maximum RoM of this new design is also more consistent across wearers of different sizes. The thigh cuff of the new design is split into two pieces with adjustable distance. Through this adjustment, the cable routing across the knee joint is decoupled from the cable routing across the hip joint. Therefore, wearers of different leg lengths have a consistent cable routing across their joints, thus achieving a more uniform RoM. The RoM of the new C-ALEX is tested on human subjects, and the results are shown in Section 3.2.

Another major improvement of the new C-ALEX is the pelvic cuff. Studies have shown that restricting the pelvis motion during walking changes the kinematics and reduces comfort [39], [44]. It is a challenging task to build an exoskeleton with mechanical linkages to include all the six degrees-of-freedom (DoF) motion of the pelvis. Using passive DoF will require the wearer to take care of part of the inertia and/or weight of the exoskeleton, while using active DoF will drastically increase the complexity and cost of the whole system. In the preliminary C-ALEX design, the pelvic motion of the user was also restricted due to the externally supported stationary pelvic cuff. In this new design, the pelvic cuff is attached to the wearer. There is no external connection to the pelvic cuff other than the soft Bowden cables, therefore the wearer can enjoy the full six DoF motion of the pelvis with minimal constraints. Overall, the six DoF pelvic cuff together with the joint-free design creates a natural walking experience for the wearer without any restrictions to natural DoFs.

Four cables are used in C-ALEX. Unlike the preliminary design, where bare cables

directly come down from the motors, Bowden cables are used to guide cables from the motors to the pelvic cuff to make the device appear less intrusive and more compact. After going through Bowden cables, bare cables are distributed around the pelvic cuff through a set of pulleys to minimize the friction. The cable routing from the pelvic cuff to the thigh and shank cuffs is shown in Fig. 2.1. Different from the preliminary design, the new C-ALEX is modeled as a four DoF two-link serial chain with a ball joint at the hip and a hinge joint at the knee. Fig. 2.1 shows the kinematic model of C-ALEX. The DH parameters of C-ALEX are shown in Table 2.1. Among the four DoF of C-ALEX, two are actuated by cables: q_1 hip flexion-extension and q_4 knee flexion-extension. The other two DoF are passive. Same as the preliminary design, each cable of C-ALEX is driven by an AKM43 AC servo motor and the tension in each cable is measured by a Futek load cell at the end of the cable. The joint angles are measured by a Vicon [1] optical motion capture system, which provides more accurate and reliable joint angle measurements than the IMUs used in the preliminary design. Reflective markers are attached to the pelvic, thigh and shank cuffs to allow optical tracking.

Table 2.1: DH Parameters of the Improved Design.

Link	a	d	α	θ
1	0	0	$\pi/2$	$q_1 + \pi/2$
2	0	0	$\pi/2$	$q_2 + \pi/2$
3	0	$-L_t$	$\pi/2$	$q_3 + \pi/2$
4	L_t	0	0	$q_4 - \pi/2$

L_t: thigh length; L_s: shank length.

In summary, the new C-ALEX inherits the main features of the cable-driven design:

[1]Vicon, Oxford, UK

the light-weight cuffs and the joint-free structure, and improves the RoM and gives complete DoF to the pelvic motion. Altogether, this new C-ALEX is an exoskeleton that adds minimal weight and inertia to the wearer with least constraint.

2.2 Improvements in the Controller

The Dynamical Model of the Improved C-ALEX

Different from the preliminary design, the new C-ALEX is modeled as a four DoF two-link serial chain instead of a two DoF double pendulum. The hip abduction/adduction as well as hip internal/external rotation DoF are also modeled in this new design. It is impossible to fully control all four DoFs with only four cables in C-ALEX, therefore, the controller only control the torque in two DoFs, the hip and knee flexion/extension DoF, same as the preliminary design. But this model can be used to calculate the un-intended torque applied to the un-controlled DoF. Except for the additional DoFs, other aspect of the dynamical model is similar to the preliminary design in Section 1.2. Based on this kinematic model, the dynamical equation of C-ALEX can be written as:

$$D(q)\ddot{q} + C(q,\dot{q})\dot{q} + G(q) = \tau_c - \tau_h \,. \tag{2.1}$$

where $q = [q_1, q_2, q_3, q_4]^T$ is the vector of joint angles, $D(q)$ is the inertia matrix, $C(q,\dot{q})$ is the vector of Coriolis and centripetal terms, $G(q)$ is the vector of gravity terms, τ_c is the actuation torques generated by the cables at each joint, and τ_h is the torques that C-ALEX applies to the human user. Ideally, τ_h should equal the intended assistive torque τ_a for

the subject. Given the light-weight nature of C-ALEX, the dynamical terms D and C are neglected. Therefore, the actuation torques τ_c equals to:

$$\tau_c = \tau_a + G(q)\,. \tag{2.2}$$

Under the transparent mode of C-ALEX, the the actuation torques τ_c is:

$$\tau_c = G(q)\,. \tag{2.3}$$

Improved Cable Tension Planning

After calculating the desired actuation torques τ_c, the next step is to find a set of cable tensions to produce the desired torque. As explained in Section 1.2, the relationship between the cable tensions and joint torques is formulated by the cable Jacobian matrix:

$$\tau_d = J(q)^T\, T\,. \tag{2.4}$$

where $T = [T_1, T_2, T_3, T_4]^T$ is the vector of cable tensions, and $J(q)$ is the cable Jacobian matrix found by

$$J(q) = -\frac{\partial(L_1, L_2, L_3, L_4)}{\partial(q_1, q_4)}\,, \tag{2.5}$$

where L_i is the length of each cable measured from the pelvic cuff to the respective anchor point. The cable tension T_i has to be kept between a minimum level T_{min} and a maximum level T_{max}, *i.e.*

$$T_{min} \leq T_i \leq T_{max}\,. \tag{2.6}$$

The goal of the cable tension planner is to find a solution T to Eq. (2.4) that is within the limits of Eq. (2.6). In the preliminary design, this is solved by quadratic programming using Eq. (2.4) as an equality constraint:

$$\begin{aligned} &\min f(T) = T^T T \\ &\text{s.t.} \quad J_C(q)^T T = U \quad \text{and} \quad T \in [T_{min}, T_{max}] \,. \end{aligned} \tag{2.7}$$

This method works well if there is indeed a solution to Eq. (2.4) within the limits of Eq. (2.6). However, if C-ALEX is outside its wrench feasible workspace, i.e. there is no possible solution of T that can satisfies the constraint in Eq. (2.4), the quadratic programming does not work. The controller of the preliminary C-ALEX mitigates this issue by using the solution of the last configuration in the wrench feasible workspace if C-ALEX goes outside the workspace. Then, once C-ALEX goes back to the wrench feasible workspace, a new solution to the Eq. (2.4) is found, which is generally different from the last available solution. This causes a discontinuity in the cable tension. Such discontinuities are undesirable as they will disrupt the gait of the wearer and cause instability to the cable tension controller. To solve this problem, new C-ALEX uses a different objective function for the quadratic programming problem:

$$\begin{aligned} &\min f(T) = \tau_e^T \tau_e + \mu\, T^T T \\ &\text{s.t.} \;\; T_{min} \leq T_i \leq T_{max} \,, \end{aligned} \tag{2.8}$$

where $\tau_e = J(q)^T T - \tau_d$. This new objective function has two components: the first component $\tau_e^T \tau_e$ attempts to find a solution that best satisfies Eq. (2.4), while the second

component $\mu T^T T$ tries to minimize the sum of square of all tensions if multiple solutions to Eq. (2.4) exist. μ is a coefficient to adjust the relative importance of these two components. Although this quadratic programming does not give a precise solution to Eq. (2.4), it guarantees that a solution can always be found across the RoM. Through careful adjustment of μ, the error in Eq. (2.4) can be kept to a practically small level.

Improved "Assist-as-needed" Controller

The "assist-as-needed" controller used in Section 1.2 generates a force field around the target gait pattern. The assistance that the user receives is entirely based on the spatial movement of the leg. The temporal aspect of the gait was not taking into consideration by the controller. Essentially, the assistive force pushes the ankle towards a target point on the target ankle path that is closest to the current ankle position, but this target point may not be at the proper location for the ankle to be at this instance. For example, at a specific instance during walking, the user may be at the 75% of the gait cycle, but the closest point on the target ankle path might be at 85% of the gait cycle. This mismatch between the phase of the gait cycle of the user and the phase of the gait cycle of the target point might negatively affects the gait training result. Therefore, a new method to find the target point on the target ankle path is used in the new C-ALEX controller.

When a subject is walking with C-ALEX, the assistive controller finds the current ankle position P_{ankle} by the joint angles q using the forward kinematics. The controller also keeps a record of the ankle positions during the past three steps, and find an average ankle path (green path in Fig. 2.2 (A)). Let a point P be the point on this average ankle

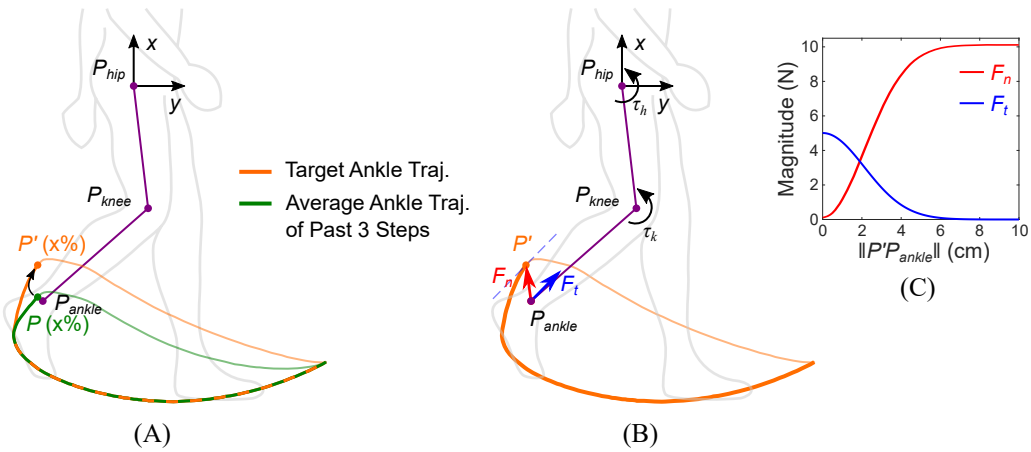


Figure 2.2: Improved "Assist-as-needed" Controller. The silhouette represents a walking subject. The purple lines are the kinematic model of the leg. P_{hip}, P_{knee} and P_{ankle} represent the locations of the hip, knee and ankle joint. (A) Find the point P on the average ankle trajectory (green) the is closest to P_{ankle}, and then find the target ankle point P' on the target ankle trajectory (orange) with the same gait phase as P. (B) Find the direction of the assistive force F_n and F_t to simulate at the ankle point. The equivalent torque of the assistive force τ_h and τ_k are generated by the cables at the hip and knee joint. (C) The magnitude of the assistive force F_n and F_t relative to the distance between P_{ankle} and P'.

path that is closest to P_{ankle}. Then, assuming that the ankle path of the current step is very close to this averaged ankle path of past three steps, the phase of point P on the average ankle path is a good estimation of the current phase of the gait. Using this phase, the controller finds the point P' on the target ankle path (orange path in Fig. 2.2(A)) at the same phase, and simulates an assistive force to push the ankle towards P'. Similar to the controller in the preliminary design, this assistive force has two components: a component F_n that is pointing to P', and a component F_t that is parallel to the tangential line at P' on the target path, as shown in Fig. 2.2 (B). The magnitude of these two forces are found in the same way as Eq. (1.16). Fig. 2.2 (C) shows how the magnitude of the two assistive forces changes as d increases. The total assistive force F_a is the sum of F_n and F_t: $F_a = F_n + F_t$, and the assistive force controller simulates this virtual assistive force

F_a by generating assistive torques at the hip and knee joints of the leg:

$$\tau_a = J_e(F_n + F_t)\,. \tag{2.9}$$

in which τ_a is the total assistive torque in Eq. (2.2) and J_e is the Jacobian matrix that relates the ankle velocity to the joint velocity. Then, the cable tension is calculated using the improved cable tension planner described in the previous subsection.

2.3 Conclusion

In conclusion, the new C-ALEX improved both its mechanical design and its controller. The newly designed cuffs of C-ALEX are light-weight, allow large RoM, and do not constrain the DoF of the user. The improved controller of C-ALEX uses a new cable tension planner to remove discontinuity in tension generation and uses a new algorithm to provide assistance based on the current phase of gait. This improved design of C-ALEX is used in the experiments detailed in the following chapters.

Chapter 3

Effects of Exoskeleton Weight and Inertia on Human Walking

Various leg exoskeletons have been designed for gait rehabilitation. The transparency of these exoskeletons is crucial to their effectiveness in gait training. The weight and inertia of an exoskeleton are two important factors affecting its transparency. The goal of this experiment is to investigate the effects of the weight and inertia of an exoskeleton on the natural walking of the subject. In this experiment, we added different levels of extra mass to C-ALEX to simulate heavier exoskeletons to study how the gait changes as exoskeleton weight and inertia change. By switching on and off the gravity compensation for the added mass, the effects of inertia can be isolated from the effects of weight.

3.1 Experiment Protocol

Twelve subjects (nine male and three female) participated in this experiment. The subjects were all healthy adults aged 22-31 with an average height of 177.2±7.5 cm and an average weight of 71.1±8.0 kg (mean±sd). The experiment was approved by the Columbia University Institutional Review Board.

Each subject first performed a RoM test during which they flexed and extended their hip and knee joints to the maximum angle that C-ALEX allowed. Then, each subject completed an experiment protocol composed of eight sessions. Each session was 5 min

treadmill walking with one particular setup of C-ALEX. At the beginning of the protocol, the subject's comfortable walking speed without C-ALEX or any added mass was determined. That same speed was used in all eight sessions of the experiment. The average walking speed of the subjects was 0.99±0.06 m/s (mean±sd).

The eight walking sessions were divided into three groups based on the setup of C-ALEX as shown in Table 3.1

(1) Free-walking group (FW): It includes three sessions with three levels of added mass: +0kg (no added mass), +1.8kg and +3.6kg. The added mass was divided evenly and attached to the thigh and shank of the subjects. The thigh and shank cuffs and all the cables of C-ALEX were removed during these sessions. This setup allowed the subjects to walk naturally without any interference from C-ALEX.

(2) C-ALEX no compensation group (NC): It includes three sessions with the same three levels of added mass as the FW group (+0kg, +1.8kg, and +3.6kg). Subjects were wearing C-ALEX in the transparent mode during these sessions, *i.e.* C-ALEX did not provide any assistance other than compensating for its own weight.

(3) C-ALEX weight compensation group (WC): It includes two sessions with +1.8kg and +3.6kg of added mass. In these sessions, C-ALEX compensates not only the weight of itself, but also the weight of the added mass.

The order of the C-ALEX setups and the weight conditions under each setup was randomized among subjects.

During the experiment, the hip and knee flexion-extension angles and ankle position relative to the hip was recorded by C-ALEX at 100Hz. Within each 5 min session, the first 180 seconds were considered as adaptation period. The data from the 180 s to the 270 s

Table 3.1: Experiment Conditions to Study the Effect of Weight and Inertia.

Groups	Sessions		
Free-walking (FW)	FW+0kg	FW+1.8kg	FW+3.6kg
C-ALEX No Comp. (NC)	NC+0kg	NC+1.8kg	NC+3.6kg
C-ALEX Wt. Comp. (WC)	N/A	WC+1.8kg	WC+3.6kg

Eight experiment sessions divided into three groups. The two boxes show the two separate repeated-measure ANOVAs used in the data analysis.

were used for data processing.

The performance of the C-ALEX controller was evaluated using the RMS torque errors during each session. The torque error τ_{err} is the difference between the desired torque τ_d and the actual torque generated by cable:

$$\tau_{err} = \tau_d - J(q)^T T_{actual} \,, \tag{3.1}$$

where $J(q)$ is the cable Jacobian and T_{actual} is the actual cable tension measured by load cells.

The gait kinematics data were divided into gait cycles at heel strike and then averaged into a single gait cycle. Four variables were used to evaluate the changes in gait kinematics: hip flexion-extension RoM (hip RoM), maximum knee flexion, step length and step height. Two separate repeated-measure analysis of variance (rmANOVA) were used to compare each kinematics measurements across the sessions. The first rmANOVA is a 3×2 design that compares the three sessions in the FW group with the three sessions in the NC group (red box in Table 3.1). This comparison shows the effect of C-ALEX itself and the combined effect of the weight and inertia of the added mass. The second

rmANOVA is a 2×2 design that compares the two sessions in the WC group with the two weighted sessions (+1.8kg and +3.6kg) in the NC group (blue box in Table 3.1). Through compensating the weight of the added mass, the effect of inertia on gait can be isolated. This comparison reveals how weight and inertia independently affect the gait the subjects.

3.2 Experiment Results

Range-of-Motion Test

Table 3.2 shows the RoM of C-ALEX compared with the RoM during the unloaded free walking (FW+0kg). The result shows that the RoM of C-ALEX is adequate for the natural walking of the subjects. No subject reported their gait being constrained by C-ALEX during the whole experiment.

Table 3.2: Range of Motion of C-ALEX in Degrees.

	Hip Flexion	Hip Extension	Knee Flexion
Inside C-ALEX [a]	43.8 ± 7.4	11.6 ± 1.8	84.3 ± 7.6
Natural Walking [b]	31.2 ± 3.7	8.1 ± 1.9	68.0 ± 6.3

[a] Maximum angle (mean±std) reachable inside C-ALEX
[b] RoM (mean±std) during natural walking outside C-ALEX

Controller Performance

Fig. 3.1 shows the RMS torque tracking error averaged across the subjects during the five sessions with C-ALEX attached. This error includes both the error from low-level PID controller and the error from the approximation in cable tension planner (2.8). The magnitude of the RMS torque error was similar under each experiment condition, and

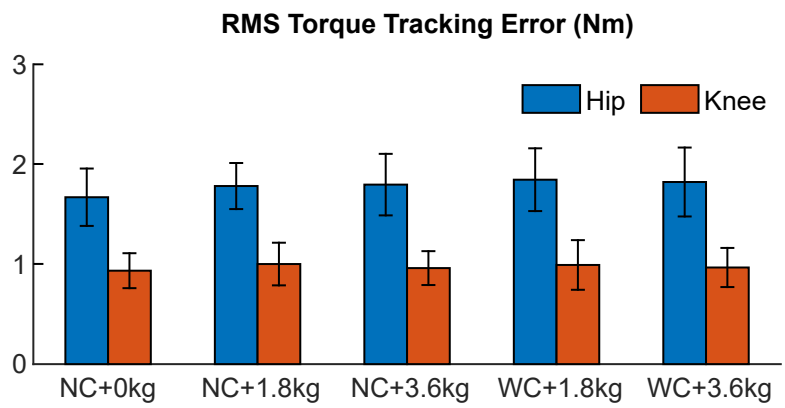


Figure 3.1: The Average RMS Torque Tracking Error during the Experiment. The vertical bars represent standard deviation.

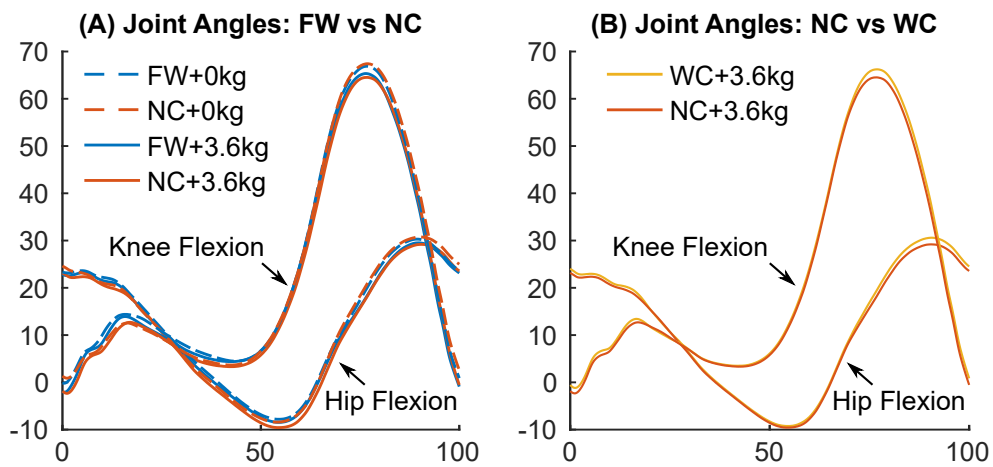


Figure 3.2: Hip and Knee Joint Angle of a Gait Cycle during Selected Sessions.

the small standard deviation bar suggests that the controller performance was consistent across all subjects.

Gait Kinematics

The joint angle trajectories averaged across the subjects during selected sessions are shown in Fig. 3.2. Fig. 3.3 shows the four kinematic variables in all sessions averaged among subjects. The rmANOVA between the FW and NC sessions shows that wearing

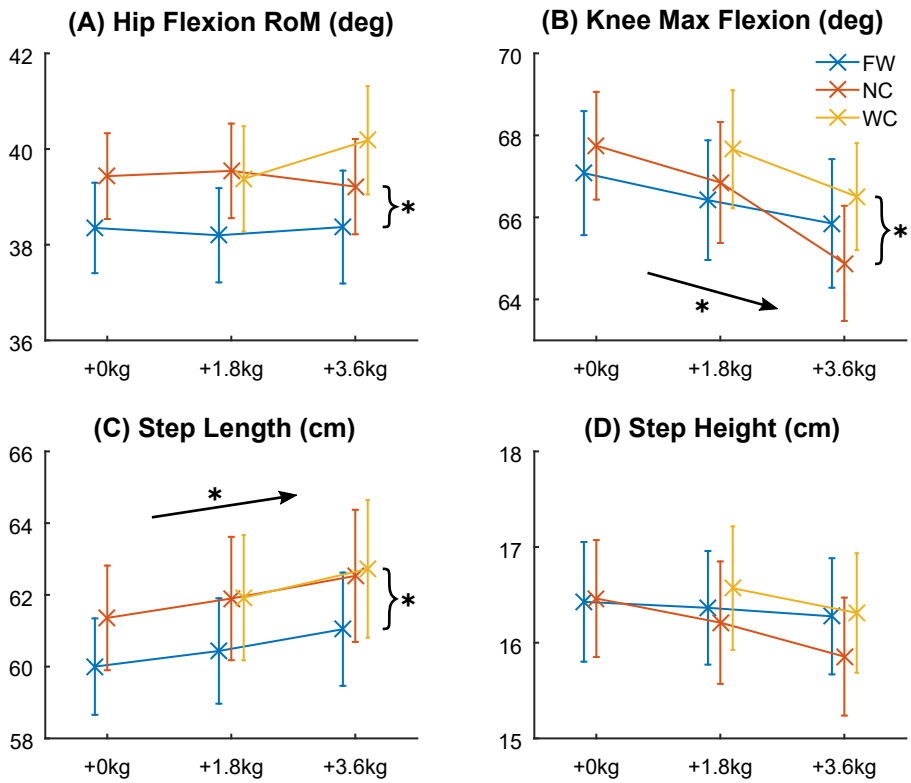


Figure 3.3: Averaged Kinematic Results from the Experiment. The vertical bars indicate the standard error. "$*$" indicates significance at 0.05 level.

C-ALEX significantly increased the hip RoM ($p = 0.004$) and step length ($p = 0.006$) of the subject. Meanwhile, increasing the added mass significantly increased the step length ($p < 0.001$) and significantly reduced max knee flexion ($p = 0.010$). The step height also reduced with added mass, but not significantly ($p = 0.074$).

The rmANOVA between the NC and WC sessions showed that compensating the weight of the added mass significantly increased hip RoM ($p = 0.043$) and max knee flexion ($p = 0.004$), while step height increased but not significantly ($p = 0.082$). An interaction between the effect of weight compensation and the effect of added mass is found in hip RoM. Post-hoc analysis within each level of added mass showed significant effect of weight compensation at +3.6kg level ($p = 0.004$), but not at +1.8kg level ($p = 0.589$).

3.3 Discussion

Effect of C-ALEX

Even though C-ALEX has minimal weight and inertia and virtually no kinematic constraint, it still caused an increase in the hip RoM and the step length. The inertia of C-ALEX itself as well as the torque tracking error shown in Fig. 3.1 could be the main factor causing this change. C-ALEX has a torque tracking error of around 2 Nm at the hip and 1 Nm at the knee, which is comparable to the result in [42]. Other factors causing the change in kinematics include possible inaccuracies in the measurement of the size of the subject and the locations of the cuffs. All these factors are consistent across the experiment conditions for the same subject, therefore they should not affect the analysis of the added weight and inertia. Overall, compared with free walking (FW+0kg), walking with C-ALEX (NC+0kg) increased the hip RoM by $1.08°$ and step length by 1.36 cm on average. The changes are relatively small compared with the results in [40], [62], showing that C-ALEX can better preserve the natural gait. It is also worth noting that C-ALEX can still benefit from more sophisticated controllers such as in [42], [63] to further improve its transparency, while it is a bigger challenge for a heavy exoskeleton to reduce its weight and inertia.

Effect of Added Mass

Increasing the level of added mass in FW and NC sessions affects all the gait variables except for the hip RoM. This proves our hypothesis that the changes in the gait of the subjects increase as the amount of added mass increase. The increase in step length due to

the added mass implies a similar increase in stride time and a decrease in cadence, giving the constant treadmill speed. Similar changes in gait have been reported by previous research when adding a simple weight to the leg [64]–[66]. In particular, both Browning *et al.* and Holt *et al.* used a series of increasing weights and found an increasing trend in step length and stride time. The change in knee flexion and step height due to added mass was reported in [64], [67] as well. All these changes come from the combined effect of weight and inertia of the added mass. No significant interaction is found between the level of added mass and the C-ALEX setup, suggesting that the added mass affects the free-walking and walking with C-ALEX in a similar way.

Effect of Weight and Inertia

The most interesting part of the result is the distinction between the effect of weight and the effect of inertia. Fig. 3.3 shows that compensating the weight of the added mass increased step height, but had almost no effect on the step length. This implies that the change in the step length is mostly due to the inertia, while the change in the step height is mostly affected by the weight. Browning *et al.* reported that moving the same amount of mass more distally (same weight but more inertia) increased step length [64], which also showed the effect of inertia on step length. The increase in step length with more inertia contradicts the result in [68]. The difference might come from how the inertia load is applied. In [68], the inertia load was purely in the anterior-posterior direction on the ankle, whereas in our experiment, the inertia load was distributed between thigh and shank and moved together with the leg segments. Our setup better simulates the inertia

load of an exoskeleton.

Besides reducing the step height, the weight of the added mass also plays a major role in the max knee flexion, as it increased significantly after weight compensation. The cause of this might be that the weight of the added mass generates a negative torque against the knee flexion, and therefore compensating for this weight allows the knee to flex more. Compensating for the weight also increases the hip RoM significantly in the session with the largest load (+3.6kg). Given that the gravity helps to slow down the forward swing of the leg, compensating for it may result in greater hip flexion and therefore an increased RoM. This theory is supported by a paired t-test to compare the maximum hip flexion between the NC+3.6kg session and the WC+3.6kg session. The result showed that weight compensation significantly increased ($p = 0.033$) the maximum hip flexion by an average of $1.36°$, which accounted for the majority of the changes in the hip RoM. Overall, weight compensation reduced the changes in max knee flexion and step height due to added mass, but also caused a greater change in hip RoM. This result suggests that only compensating for the weight of the exoskeleton may negatively affect the gait in some aspects.

Limitations

Overall, the changes in gait kinematics observed in this experiment were small. The reason might be that the amount of the added mass was limited. Studies have shown that healthy subjects were able to preserve the gait pattern under small loads at the expense of increased muscle activity and metabolic cost[64], [65], [68]. This experiment did not measure muscle activity and metabolic cost, but one would expect a similar increase. El-

der and motor impaired population usually have reduced lower limb strength and are more sensitive to external perturbation [69], therefore, the effect of exoskeleton weight and inertia are likely to be more prominent. In the future, this experiment needs to be validated with greater added mass, include more measurements and expand the subject populations.

3.4 Conclusion

In summary, this study discovered that both weight and inertia of an exoskeleton affect the gait of the user. Compensating for the weight alone partially restored some aspects of the gait, but is not enough to bring the gait back to natural, as certain aspects of the gait are almost exclusively affected by the inertia. The amount of added mass used in this experiment is relatively small. The effects of weight and inertia are likely to be more prominent on a large exoskeleton, thereby requiring complicated sensing and control strategy to improve transparency. This result shows the benefit of a light-weight exoskeleton such as C-ALEX, which can better preserve the natural gait of the subject without additional complicated efforts in sensing and control.

Chapter 4

Gait Training with C-ALEX on Healthy Subjects

The experiment in Section 1.3 showed the effectiveness of C-ALEX in gait assistance. Subjects were able to better follow a new gait pattern with the help from C-ALEX. However, as the C-ALEX is designed for training and rehabilitation, it is important to show that the subjects can still retain the new gait pattern after taking off C-ALEX. This chapter investigated the effectiveness of C-ALEX in gait training on a group of health individuals.

4.1 Experiment Protocol

Nine healthy young adults participated in this study. The subjects were aged between 22 and 32 with an average height of 1.76 $\pm$ 0.07 m and an average weight of 69.7 $\pm$ 8.1 kg. The goal of the experiment was to train the subjects to walk in such a way that their ankle point follows a specific target path in the sagittal plane. This target ankle path was modified from the average ankle path of the baseline gait of the subject recorded during the free-walking session with a 30% increase in step height, as shown in Fig. 4.2 (A). The stance phase of the baseline ankle path is unchanged, while each point of the swing phase of the baseline path is moved up vertically by Δy based on the horizontal position of the point. The relationship between Δy and the horizontal position is shown in Fig. 4.2 (B).

Fig. 4.3 shows the training protocol for the study. At the beginning of each experi-

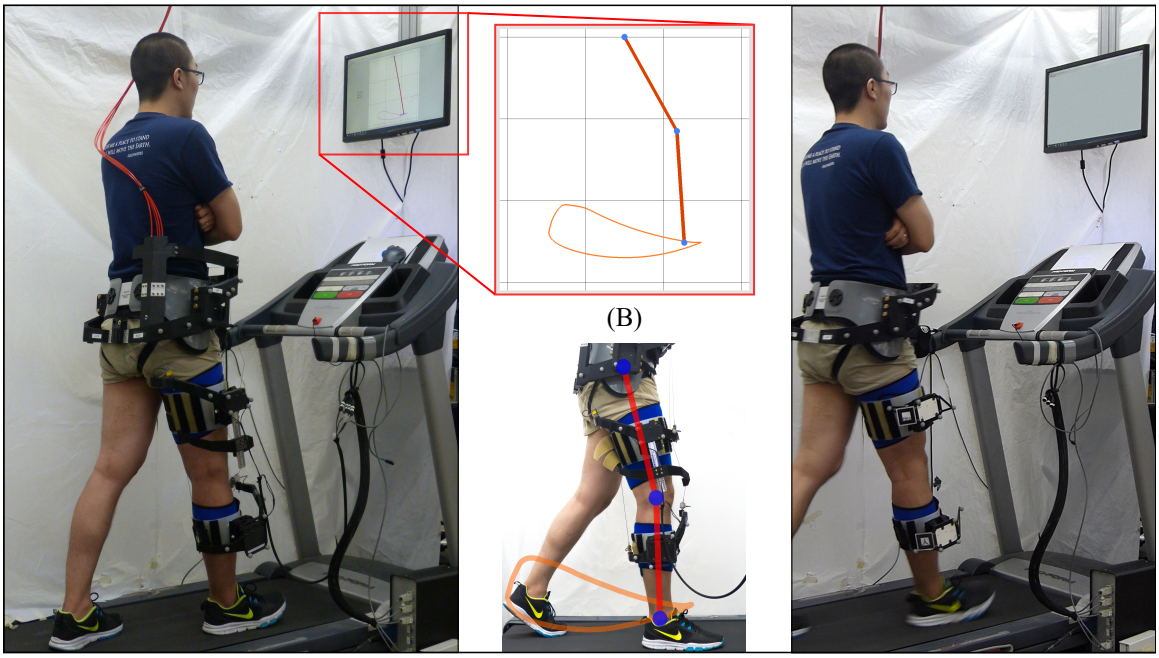


Figure 4.1: The Experiment Setup for Gait Training Experiment with C-ALEX. (A) The experiment setup during the training sessions. The subject wears C-ALEX with real-time visual feedback of gait. (B) Visual feedback displayed in front of the subject during the experiment. The red line represents the leg; the blue circle represents the hip, knee and ankle joints; the orange line is the target ankle path. (C) This shows how the visual feedback correlates to the leg of the user. (D) The experiment setup when C-ALEX was not worn. The subject only wore a light-weight cuffs for motion tracking purpose. This setup is used in the free-walking session and the post-training sessions to record how the subject works without wearing C-ALEX.

ment before putting on C-ALEX, comfortable walking speed of each subject was found and the same speed was used in all experiment sessions. The average walking speed of the subjects was 3.40 ± 0.13 km/h. Then, each subject walked without C-ALEX for 3 minutes as a free-walking session (FW) to capture the baseline gait (Fig. 4.1 (D)). After this free-walking session, C-ALEX was attached to the subject and the subject completed a 3-minute pre-training session (PreT) to be acclimated to walking with C-ALEX. During this pre-training session, C-ALEX force assistance was turned off. After the pre-training

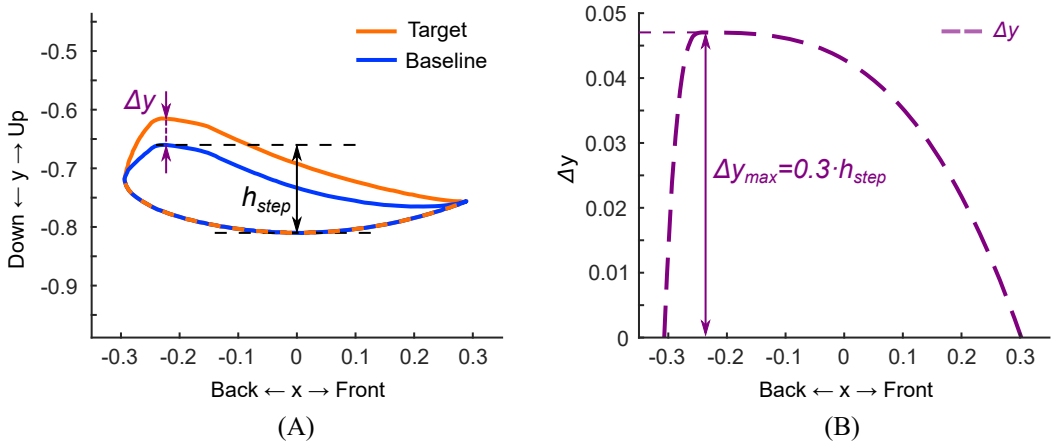


Figure 4.2: Generating Target Ankle Trajectory for Gait Training with C-ALEX. (A) The target path generated from the baseline path. The stance phase is the same while the swing phase was increased by Δy. (B) The relationship between Δy and the horizontal position.

session, the training started.

There were four training sessions (TR-1~4) to train the subjects to follow the target ankle path. Each training session was 10 minutes of continuous walking. C-ALEX force assistance was switched on during the entire training sessions. During each training session, visual feedback was provided intermittently. As shown in Fig. 4.1 (B), the visual feedback shows the current position of the ankle and the target ankle path. Fig. 4.1 (A) shows training session setup when the subject was walking with the visual feedback displayed in the front. At the end of each training session, subjects continued walking for another 30 seconds. The assistance of C-ALEX was turned off without the subject's knowledge. This 30 seconds of walking after each training session are considered as a catch trial (CT-1~4). The subjects stopped walking after the catch trial and took a one-minute break. After the break, the subject walked for another minute with C-ALEX assistance turned off. This one-minute session is used as a mid-training session to see how much they have

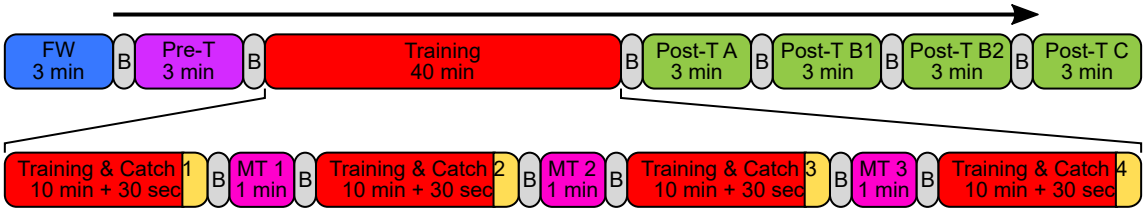


Figure 4.3: Experiment Protocol for Gait Training with C-ALEX. 40 minutes of training is further divided into four 10-minute training sessions, each with a catch sessions at the end, FW: Free-walking Session; Pre-T: Pre-training Session; MT: Mid-training Session; Post-T: Post-training Session; B: Break

learned so far. There are three mid-training sessions (MT-1~3) in total between the four training sessions. After the mid-training sessions, the subjects took a 5-minute break before starting the next training session.

After the last training session, there are four post-training sessions (PT-A, PT-B1, PT-B2, PT-C in this order) detailed in Table 4.1 to evaluate their learning performance. Breaks were provided between these sessions. During the PT-A, C-ALEX remained on the subjects but did not provide any assistance. Then the subjects took off C-ALEX and completed the remaining sessions (PT-B1, PT-B2 and PT-C). During PT-A, PT-B1 and PT-B2, the subjects were instructed to try their best to keep walking the way they were trained, while in PT-C, the subjects were instructed to ignore the training and not think about how they walk. Since the subjects had been focusing on their gait since the beginning of the training, it might be hard for them to not to think about it. Therefore, we displayed a video of cute cats [70] in front of the subjects to distract them.

Table 4.1: Conditions of the Post-training Sessions

Session	Wear C-ALEX	Instruction	Start Time	Duration
PT-A	Yes	Walk as trained	1 min after training	3 min
PT-B1	No	Walk as trained	10 min after training	3 min
PT-B2	No	Walk as trained	18 min after training	3 min
PT-C	No	Ignore the training	22 min after training	3 min

4.2 Data Processing and Statistical Analysis

During the experiment, the C-ALEX controller calculates the ankle position relative to the hip joint in real-time based on the joint angles tracking data from the motion capture system. This ankle position is used to find the assistance force in real-time, and is also saved for later data analysis.

Table 4.2: Time from Each Session Used in Data Analysis

Session	FW	PreT	TR-1~4	CT-1~4	MT-1~3	PT-A~C
Time	100s~160s	100s~160s	510s~570s	605s~620s	15s~45s	100s~160s

For each session of the experiment, a portion of the ankle position data in the middle of the experiment is used in the data analysis. The exact portion used is shown in Table 4.2. From the data in this portion, the ankle path in the sagittal plane was divided into gait cycles at the approximate heel-strike event, which is determined when the ankle reaches the forward most position in each gait cycle. Then the ankle path data within each cycle was resampled to the same length of 500 data points at a constant time interval. The resampled ankle path was averaged into a single gait cycle as the average ankle path of that session. The step height of the session is calculated as the difference between the maximum and minimum vertical value of the averaged ankle path. The deviation area is

calculated as the non-overlapping area between the averaged ankle path and the target ankle path.

The subjects were instructed to not only reach the increased step height, but also follow the exact ankle path of the target gait. Two metrics were used to evaluate the performance of the subjects in following the target ankle path. The first one, normalized step height, is the step height of the current session divided by the step height of the baseline gait of the subject. Therefore, the normalized step height of the baseline gait is 1.0, while the normalized step height of the target gait is 1.3. The second metric, normalized tracking error, shows the amount of error in following the ankle path. The tracking error is measured by the deviation area between the average ankle path of the current session and the target ankle path (shaded area of Fig. 4.4 (B)), and it is then normalized by the deviation area of the baseline gait (shaded area of Fig. 4.4 (A)). A lower normalized tracking error suggests that the subjects can follow the target gait better. If a subject could follow the target gait perfectly, the normalized tracking error is 0. As the subject deviates away from the target path, the normalized tracking error increases. If a subject completely ignored the training and simply walked in their baseline gait, the normalized tracking error of the baseline gait is 1.0. Both of these metrics has been used in previous researches [39], [71]–[74].

From PreT to PT-C, there are 16 sessions in total. The two gait metrics of each session averaged among subjects were compared against the baseline value of 1.0 using one-sample t-test. To correct for multiple comparisons, Bonferroni-Holm correction was used to adjust the p-values. Selected results from the one-sample t-test were reported in the Result section. To study whether the gait performance of the subjects changed during

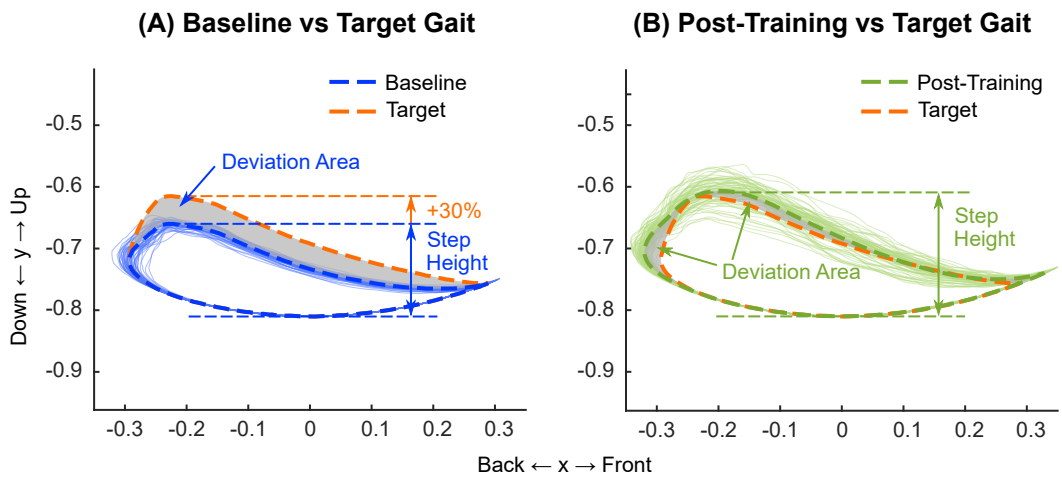


Figure 4.4: Gait Training Effectiveness Metrics. (A) Target gait generated from baseline gait. The thin blue lines are the ankle path of the baseline gait of a subject, while the bold dashed blue line is the averaged ankle path of the baseline gait. The shaded area is the deviation area between the baseline gait and target gait. (B) Post-training gait compared with the target gait. The thin green lines are the ankle path of the post-training session, while the bold dashed green line is the averaged ankle path of the baseline gait.

the four post-training sessions, a one-way repeated-measure analysis of variance (rmANOVA) was used on each of the two gait metrics, with post-hoc pair-wise comparison adjusted by Bonferroni correction. The same statistical procedure was also used to compare the two gait metrics between the MT-1, MT-2, MT-3 and PT-A sessions to study the training progress. To compare the two pairs of training sessions and catch sessions, a 2×2 two-way rmANOVA was used. In all statistical analysis, $\alpha = 0.05$.

4.3 Experiment Result

Effects of Training: Results from the Post-training Sessions

There are four post-training sessions. Post-training session A (PT-A) started 1 minute after the training. In this session, the subjects were still wearing C-ALEX, but C-ALEX

was not assisting the subjects. Post-training sessions B1, B2 and C (PT-B1, PT-B2 and PT-C) started 10, 18 and 22 minutes after the training, respectively. During these sessions, the subjects took off C-ALEX and walked without the exoskeleton, just like they did during the free-walking sessions. During PT-A, PT-B1 and PT-B2, the subjects were instructed to try their best to repeat the target gait pattern, however, during PT-C, the subjects were instructed to ignore the training and not to think about how they walk. A video distraction was displayed to the subjects to divert their attention. Fig. 4.5 shows the ankle trajectories of a representative subject during the four post-training sessions. The normalized step height and normalized tracking error averaged among all subjects during the post-training sessions are shown in Fig. 4.6 (A).

During PT-A, the average normalized step height was 1.22 ± 0.12 (mean±std), close to the target of 1.30, and significantly greater than the baseline ($p = 0.002$). The ankle path of this session was close to the target ankle path. The normalized tracking error averaged among subjects was 0.54 ± 0.17 (mean±std), significantly less than the baseline of 1.00 ($p < 0.001$).

During PT-B1 and PT-B2, taking off C-ALEX did not affect the retention of the target gait. The normalized step height averaged among subjects were 1.29 ± 0.13 and 1.31 ± 0.15 respectively (mean±std), close to the target of 1.30, and significantly greater than the baseline ($p_{B1} = 0.003$, $p_{B2} = 0.004$). The normalized tracking error of these two sessions were 0.60 ± 0.17 and 0.58 ± 0.19 (mean±std) respectively, significantly less than the baseline ($p_B1 = 0.002$, $p_B2 = 0.002$).

During PT-C, despite the instruction to ignore the training, the gait of the subjects did not go back to their baseline. The normalized step height was 1.11 ± 0.08 (mean±std)

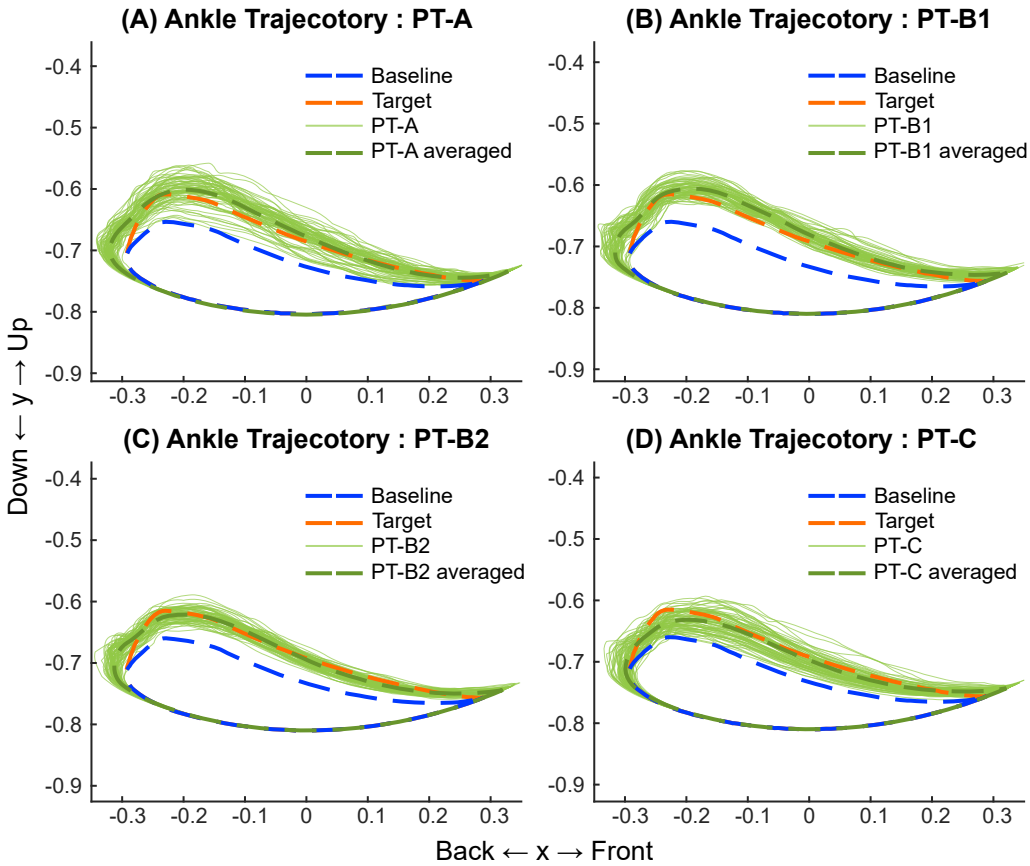


Figure 4.5: Ankle Trajectories during the Post-training Sessions. The figure shows the ankle trajectories during the post-training sessions of a representative subject. The thin green lines are the ankle trajectories during the experiment and the thick dashed green line is the averaged ankle path of the session. The blue and orange dashed lines are baseline and target trajectories for comparison.

during this session. Although it was not as close to the target of 1.30 as the other post-training sessions were, it was significantly higher than the baseline gait ($p = 0.008$). The normalized tracking error was 0.73 ± 0.40 (mean±std). It was less that the baseline, but no significance was found ($p = 0.080$). Comparison among the post-training sessions found significant difference in normalized step height ($p < 0.001$), and pairwise comparison found that the normalized step height of PT-C is significantly less that PT-A ($p = 0.036$), PT-B1 ($p = 0.050$) and PT-B2 ($p = 0.034$). No difference in normalized tracking error is

found ($p = 0.256$).

Overall, the results from the post-training sessions showed that the gait of the subjects after training was significantly different from their baseline gait and became closer to the target gait. This suggests that the subjects were able to retain the trained gait pattern after training, which indicates that training with C-ALEX was effective.

The Progress of Training: Results from the Mid-Training Sessions

The progress of the training can be observed from the mid-training sessions (MT-1~3) and PT-A. These sessions are all done 1 minute after a training session. The pre-training sessions (PreT) are also added for comparison. During all these sessions, the subjects were wearing C-ALEX but did not receive any gait assistance. The result is shown in Fig. 4.6 (B).

During the PreT sessions, the subjects were adapting to walk inside C-ALEX, and they had not received training yet. Therefore, the gait is similar to the baseline. The normalized step height decreased to 0.96 ± 0.05 (mean±std) but was not significantly different from baseline ($p = 0.059$). The normalized tracking error increased to 1.21 ± 0.24 (mean±std), also not significantly different from the baseline ($p = 0.094$).

The normalized step height of the mid-training sessions were 1.26 ± 0.14, 1.31 ± 0.17 and 1.26 ± 0.14 (mean±std) respectively, close to the target of 1.30 and all significantly greater than the baseline ($p_{MT1} = 0.006$, $p_{MT2} = 0.006$, $p_{MT3} = 0.006$). The normalized tracking error of the mid-training sessions are 0.62 ± 0.25, 0.64 ± 0.29 and 0.62 ± 0.31 (mean±std) respectively, all significantly less than the baseline ($p_{MT1} = 0.015$,

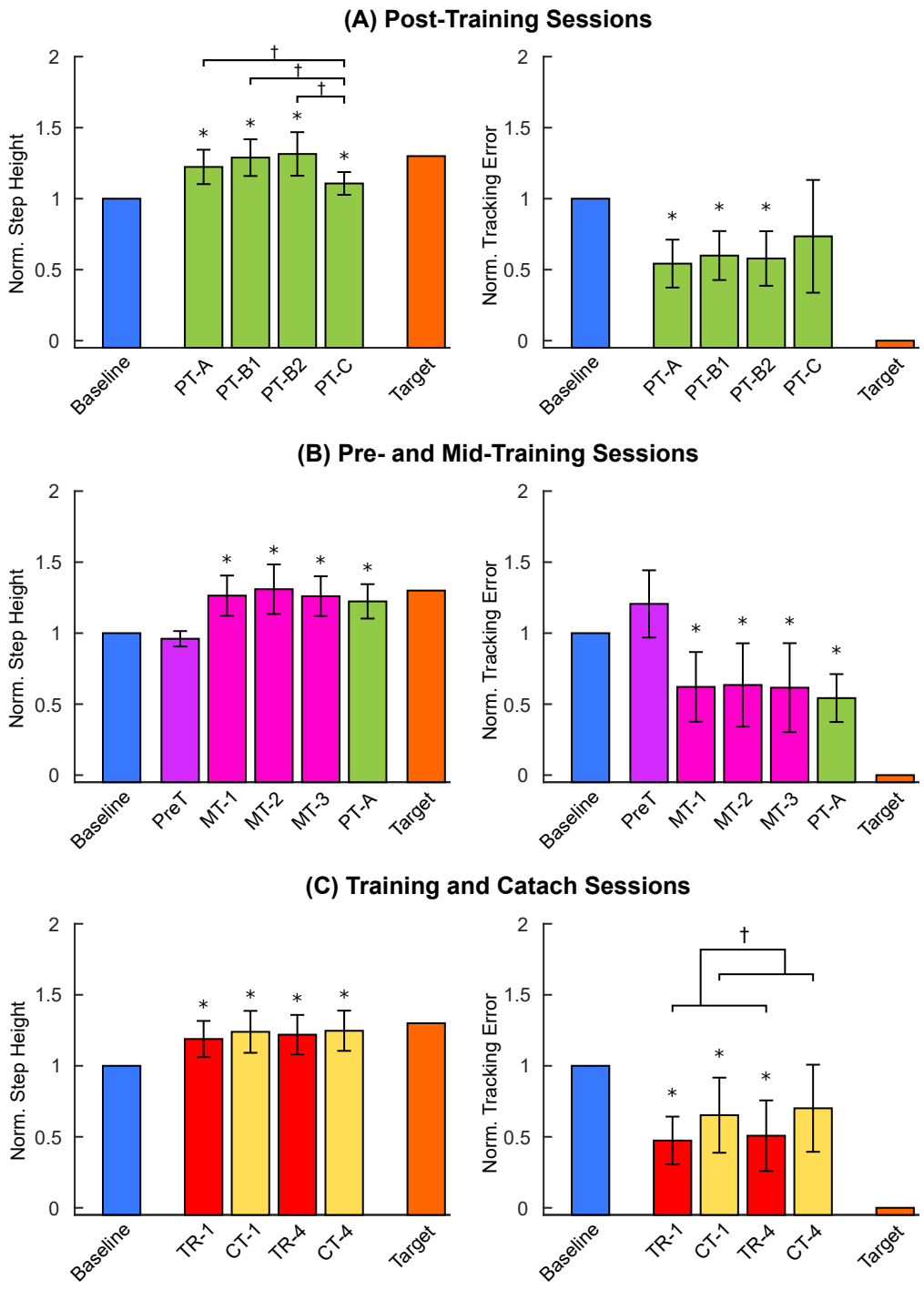


Figure 4.6: Kinematic Results from Selected Sessions of the Training Experiment. The figure shows the normalized step height and deviation area averaged among subjects during selected sessions. The baseline and target value is also shown for comparison. The error bars denote standard deviation. "$*$" denote significant difference from the baseline at $p < 0.05$ level. Pairwise comparison are done among the post-training sessions. "†" denotes significant difference between these sessions at $p < 0.05$ level.

$p_{MT2} = 0.035$, $p_{MT3} = 0.031$). Comparison among the MT-1~3 and PT-A sessions found no significant difference in neither the normalized step height ($p = 0.228$) nor the normalized tracking error ($p = 0.687$).

Effect of Force Assistance during Gait Training

How much the subjects relied on the force assistance to follow the target gait during training can be observed from the catch sessions. The catch session is the last 30 seconds of each training session during which the force assistance is turned off without the knowledge of the subject. The result from the first training and catch session (TR-1 and CT-1) as well as the last training and catch session (TR-4 and CT-4) is shown in Fig. 4.6 (C).

During TR-1 and TR-4, the normalized step height are 1.19 ± 0.13 and 1.22 ± 0.14 (mean±std), respectively. Both are significant greater than the baseline ($p_{TR1} = 0.006$, $p_{TR4} = 0.007$). The normalized tracking error of TR-1 and TR-4 are 0.47 ± 0.17 and 0.51 ± 0.25 (mean±std), respectively. Both are significantly less than the baseline ($p_{TR1} < 0.001$, $p_{TR4} = 0.003$). During CT-1 and CT-4, the normalized step height are 1.24 ± 0.15 and 1.25 ± 0.14 (mean±std), respectively, significantly greater than the baseline ($p_{TR1} = 0.006$, $p_{TR4} = 0.007$). The normalized tracking error of CT-1 is 0.65 ± 0.26, significantly less than the baseline ($p_{TR1} < 0.001$). The normalized tracking error of CT-4 is 0.70 ± 0.31 (mean±std), no significance is found comparing to the baseline ($p_{TR4} = 0.077$).

Comparing between the training sessions and catch sessions found that the normalized tracking error is significantly less during the training sessions than during the catch sessions ($p = 0.023$), suggesting that the force assistance is effectively helping the sub-

jects to follow the target gait. No significance is found in the normalized step height ($p = 0.358$).

4.4 Discussion

The experiment results showed that the subjects were able to adapt to the new gait through training with C-ALEX, and retained the new gait during the post-training sessions. The step height during the post-training sessions increased significantly compared to the baseline gait, and was very close (within ± 6 %) to the step height of the target gait. The ankle path during the post-training sessions became significantly closer to the target path, with the tracking error reduced 40 % to 46 % compared to the baseline gait before training. Gait training studies on healthy subjects with ALEX series exoskeletons [39], [71], [72] as well as Lokomat [73] all reported 40 % to 50 % reduction in the tracking error after the training. This suggests that training with C-ALEX is as effective as the rigid-link based exoskeletons.

An important aspect of the results with C-ALEX is that the subjects took off C-ALEX during the post-training sessions and were still able to repeat the trained gait. As previous studies have shown, putting on an exoskeleton would change the dynamics of the leg and alter the gait of the wearer [62], [67], [75]. Likewise, taking off the exoskeleton also alters the dynamics of the leg, and one would expect the gait of the user to change as well. Indeed, the result of this study showed that the step height reduced as the subjects put on C-ALEX (PreT compared with FW); and increased after the subjects took off C-ALEX (PT-B1 compared with PT-A). However, the changes in step height were small and not sig-

nificant. The performance of the subjects during the PT-B1 and PT-B2 sessions are largely the same as in the PT-A session. Therefore, taking off C-ALEX did not significantly affect the gait retention after training, likely due to its lightweight and non-restrictive nature. For a rigid-link based exoskeleton with greater inertia and more movement restrictions, removing the exoskeleton may cause greater changes in gait and negatively affect the gait retention. However, few gait-training studies included a post-training session with the exoskeleton taken off, as the exoskeleton usually also served as a gait-measuring device in these studies.

The post-training session C shows the ability of subjects to walk in the trained gait with minimal cognitive effort. During the PT-A, B1 and B2 sessions, the subjects were consciously focusing on their gait to repeat the trained gait pattern. However, everyday walking is generally an automatic activity that involves little cognitive effort [76]. The purpose of the PT-C session was to investigate whether the training effect persist when the subjects were not focusing on their gait. To minimize the cognitive involvement, we instructed the subjects not to think about the training or how they walk, but focus on a video displayed in front of them to divert their attention. The performance of the PT-C session was not as good as the other three post-training sessions, with their step height reduced and tracking error increased. This shows that repeating the trained gait pattern still requires cognitive effort. It is also important to factor that PT-C session was performed 22 minutes after training and healthy subjects will slowly dissipate the effects of the training. However, the gait of the PT-C session was also significantly different from the baseline gait. This suggests that the subjects were not solely relying on their cognitive ability to repeat the trained gait pattern, and the training affected the walking

motor pattern of the subjects. One issue with this experiment is that watching a video may not be cognitive engaged enough to fully divert the attention of the subjects from their gait. This experiment should be repeated in the future using other attention distracting techniques such as counting backwards [77].

The mid-training sessions showed no change in performance as training progressed. The performance of the MT-1 sessions after 10 minutes of training is similar to the PT-A after 40 minutes of training in terms of step height and tracking error. This suggests that after one training session of 10 minutes, the subjects were able to repeat the trained gait well, and longer training time did not improve the performance. However, all the mid-training sessions were done immediately after the training. While the longer training did not improve the performance, it is possible that it helped the subjects to consolidate the learned gait pattern and retain it for a longer period. With 40 minutes of training, the subjects were able to repeat the new gait more than 20 minutes after training without significant reduction in performance, as shown by PT-B2 session. This suggests that the training effect might last for a longer time.

There are several limitations of the current study. In this study, we only collected kinematic data but did not record physiological measurements, such as electromyography (EMG) and metabolic costs. It has been shown that an exoskeleton can cause changes in the muscle activity and metabolic cost due to its weight and inertia [62], its restriction on the user's motion [45], and its controller [78]. Whether this is the case with the lightweight and restriction-free C-ALEX and its "assist-as-needed" controller requires further studies. The current study is a single session study with healthy subjects. Stroke patients training usually involves multiple sessions over longer period of time [20]. One

still needs to evaluate this system on stroke patients to determine the effectiveness of gait retraining in this population. However, previous studies have shown that exoskeletons which are effective in gait retraining of healthy individuals are also effective in improving the gait of stroke patients [6], [20]. Therefore, we expect C-ALEX to be effective in the stroke population, and we will conduct further experiments to verify this. Lightweight cable-driven exoskeletons have the potential to be incorporated in body suits and regular clothes can be worn by users. Although C-ALEX has a compact profile on the body of the user, it still requires a motion capture system on a fixed frame to monitor the leg joint angles. Future design effort on C-ALEX will focus on integrating wearable IMU sensor and using smaller actuators to make a fully wearable design.

4.5 Conclusion

Overall, the study shows that healthy subjects can learn to walk in a new gait after training with C-ALEX for 40 minutes. The training results with C-ALEX are comparable to rigid-link exoskeletons while the C-ALEX design is more compact, lightweight, and potentially customizable. This study shows that C-ALEX is a good candidate to be used in robot assisted gait training of stroke patients, and this will to be investigated in future studies.

Chapter 5

Evaluation of C-ALEX on a Stroke Patient

Since C-ALEX is intended for use in stroke rehabilitation, this chapter evaluated the performance of C-ALEX on a single stroke patients. The stroke patient tested C-ALEX in a single session study. The goal of the study was to see how the gait of the stroke patient changes while walking with C-ALEX, and whether the gait of the stroke patient improves after training.

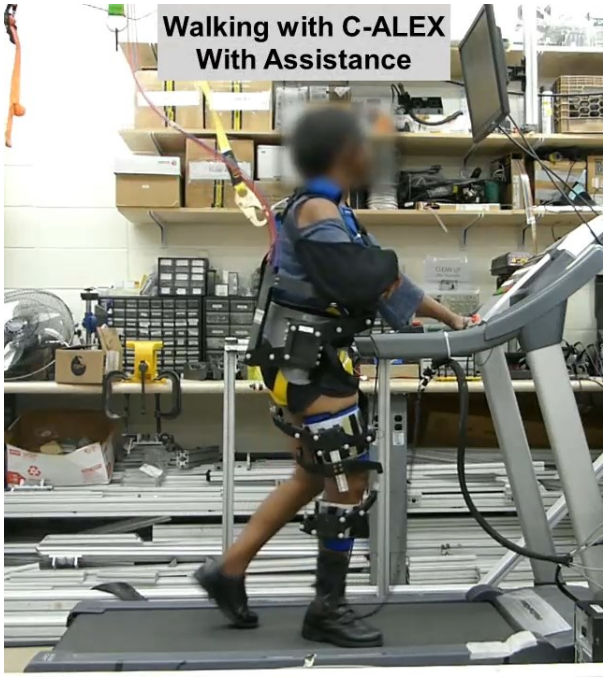


Figure 5.1: Experiment Setup for Stroke Patient Training with C-ALEX. The subject was attempting to follow a target gait pattern displayed in front of her with the help from C-ALEX.

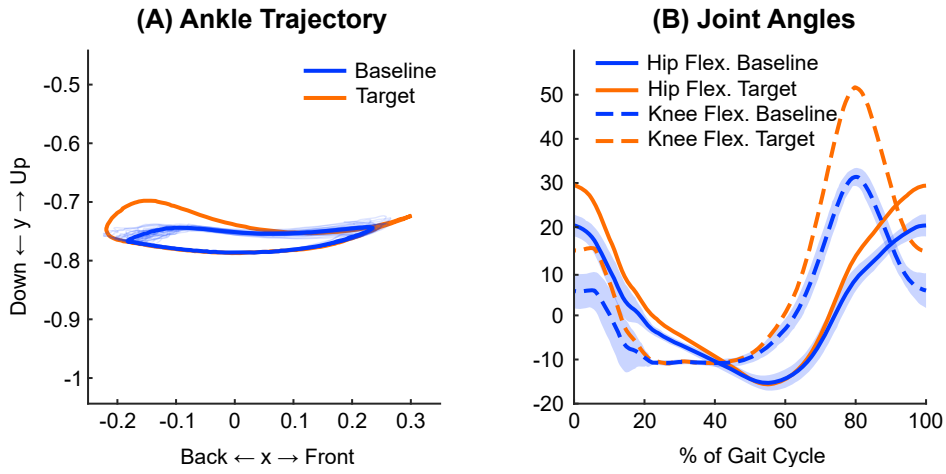


Figure 5.2: Generating Target Ankle Trajectory for the Stroke Patient. The hip and knee flexion-extenion was increased by 25% and 50% respectively to generate the target ankle trajectory from baseline trajectory.

5.1 Experiment Protocol

A single stroke patient participated in this study. The patient is 50 years old, with a height of 1.59 m and a weight of 79 kg. Similar to the training experiment of the healthy subjects, the stroke patient was trained with C-ALEX to follow a specific ankle trajectory. This target ankle trajectory was obtained from the average ankle trajectory of the baseline gait of the subject recorded during the free-walking session with an increase in the range of motion of the hip joint and knee joint. Specifically, the hip flexion-extension range of motion is increased by 25%, and the knee range of motion by 50%. The resulting target ankle trajectory is shown in Fig. 5.2.

Fig. 5.3 shows the training protocol for the stroke patient. This protocol is similar to the protocol to train healthy subjects in Chapter 4, but simplified to reduce the total experiment time and avoid fatiguing the patient. At the beginning of the experiment, the comfortable walking speed of the patient was found to be 1.29 km/h. The experiment

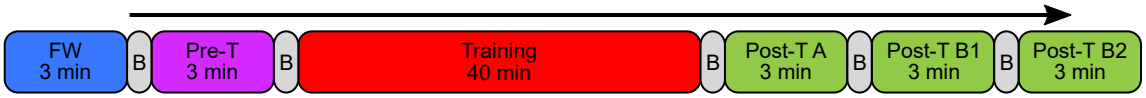


Figure 5.3: The Experiment Protocol for Training the Stroke Patient. FW: Free-walking Session; Pre-T: Pre-training Session; Post-T: Post-training Session; B: Break

started with a free-walking session (FW) without C-ALEX for 3 minutes to capture the baseline gait of the patient. Then the patient wore C-ALEX and walked for another 3 minutes without force assistance as a pre-training session (Pre-T). After the pre-training session, the training started. The training with C-ALEX lasted for 40 minutes in total. The training started with four 5-minute training sessions. As the patient became more confident walking with C-ALEX, she followed with two 10-minute training sessions. Visual feedback was provided throughout the training session. After the training sessions, the patient did three post-training sessions of 3 minutes each. In the first post-training session (PT-A), C-ALEX was still attached. In the following two post-training sessions (PT-B and PT-C), C-ALEX was removed. During PT-A and PT-B, the patient was instructed to keep walking as trained, while during PT-C, the patient was instructed to walk freely. Breaks were provided in between each session.

5.2 Data Processing and Statistical Analysis

During the experiment, C-ALEX recorded the kinematics of the patient including the hip and knee joint angles, as well as the ankle position relative to the hip joint. These data were saved for later data analysis.

A portion of the kinematic data in the middle of each session was used for the data

analysis. For the FW, PreT and PT sessions, the data from 60s to 120s were used. For the training sessions, the data during the 16th minute, 26th minute and 36th minute of training were used in the data analysis, and they are denoted as TR-16, TR-26, and TR-36, respectively. Several gait parameters during these portions were divided into gait cycles at the approximate heel-strike events, determined as the anterior-most position of the ankle trajectory in each gait cycle. The kinematic measurements of each gait cycle are averaged and reported. These gait parameters include step length and step height, deviation area from the target ankle trajectory, hip flexion-extension range of motion, maximum hip abduction and maximum knee flexion. In particular, the deviation area from the target ankle trajectory was calculated using the ankle trajectory during walking projected onto the sagittal plane.

5.3 Experiment Results

Fig. 5.4 shows the ankle trajectory of patient projected on the sagittal plane during Pre-T, TR-36, PT-A and PT-B sessions. During the Pre-T session, as the patient put on C-ALEX, the ankle trajectory of the patient is still close to the baseline with an increase in step height. During TR-36, the ankle trajectory became closer to the target trajectory after 35 minutes of training. During the PT-A session, the ankle trajectory was still close to the target trajectory without the force assistance from C-ALEX. After the patient removed C-ALEX during the PT-B session, the gait of the patient became deviated from the target gait, but still distinct from the baseline gait.

The gait parameters during the sessions are shown in Fig. 5.5. The gait parameters

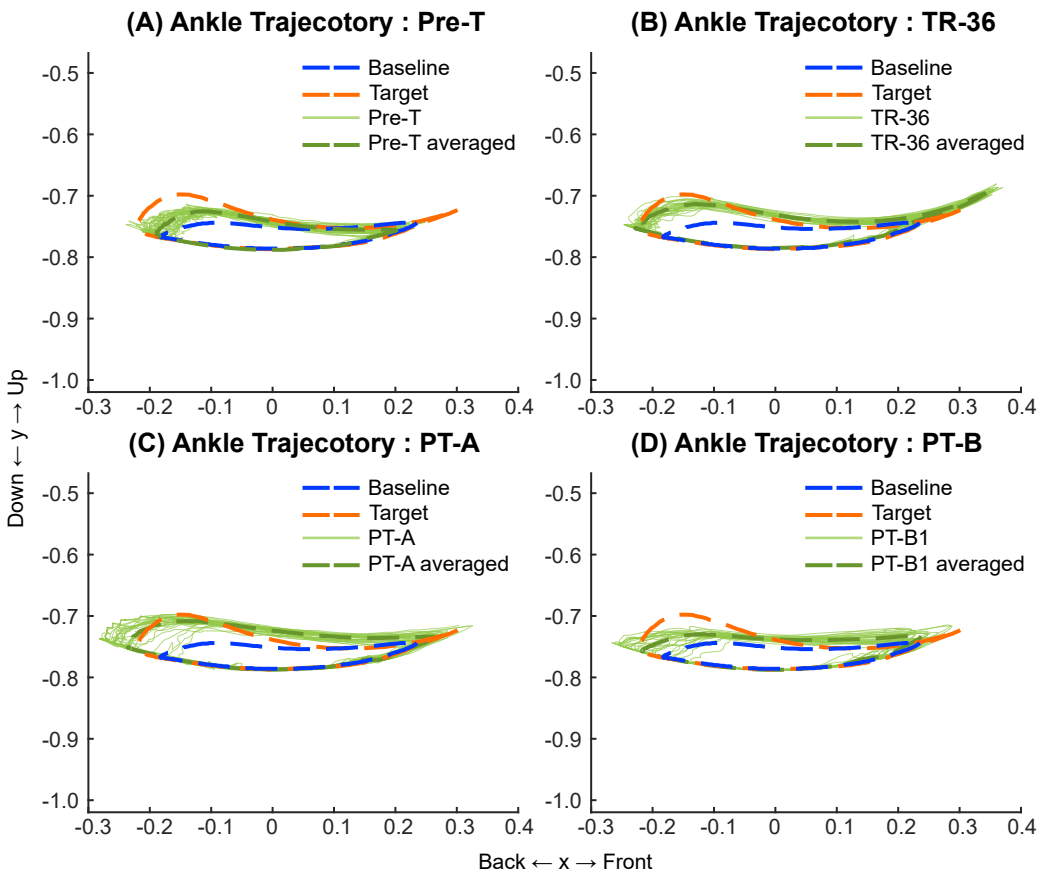


Figure 5.4: Ankle Trajectories of the Stroke Patient during Selected Sessions. The thin green lines are the ankle trajectories during the session, and the thick dashed green line is the averaged ankle trajectory of the session. The blue and orange dashed lines are baseline and target trajectories for comparison.

of the Pre-T session was close to the baseline gait, except for increased step height and reduced deviation area. During the training sessions (TR-16, TR-26, and TR-36), both the step length and step height increased from the baseline gait, and the gait was closer to the target gait with a smaller deviation area. In the joint space, the hip flexion-extension RoM, maximum hip abduction, and max knee flexion all increased during the training. In the PT-A session, there is an increase in step length and a decrease in step height compared with the gait during the training, but both are larger than the baseline gait. The deviation area increased compared to training and became closer to baseline. In the PT-B and PT-C

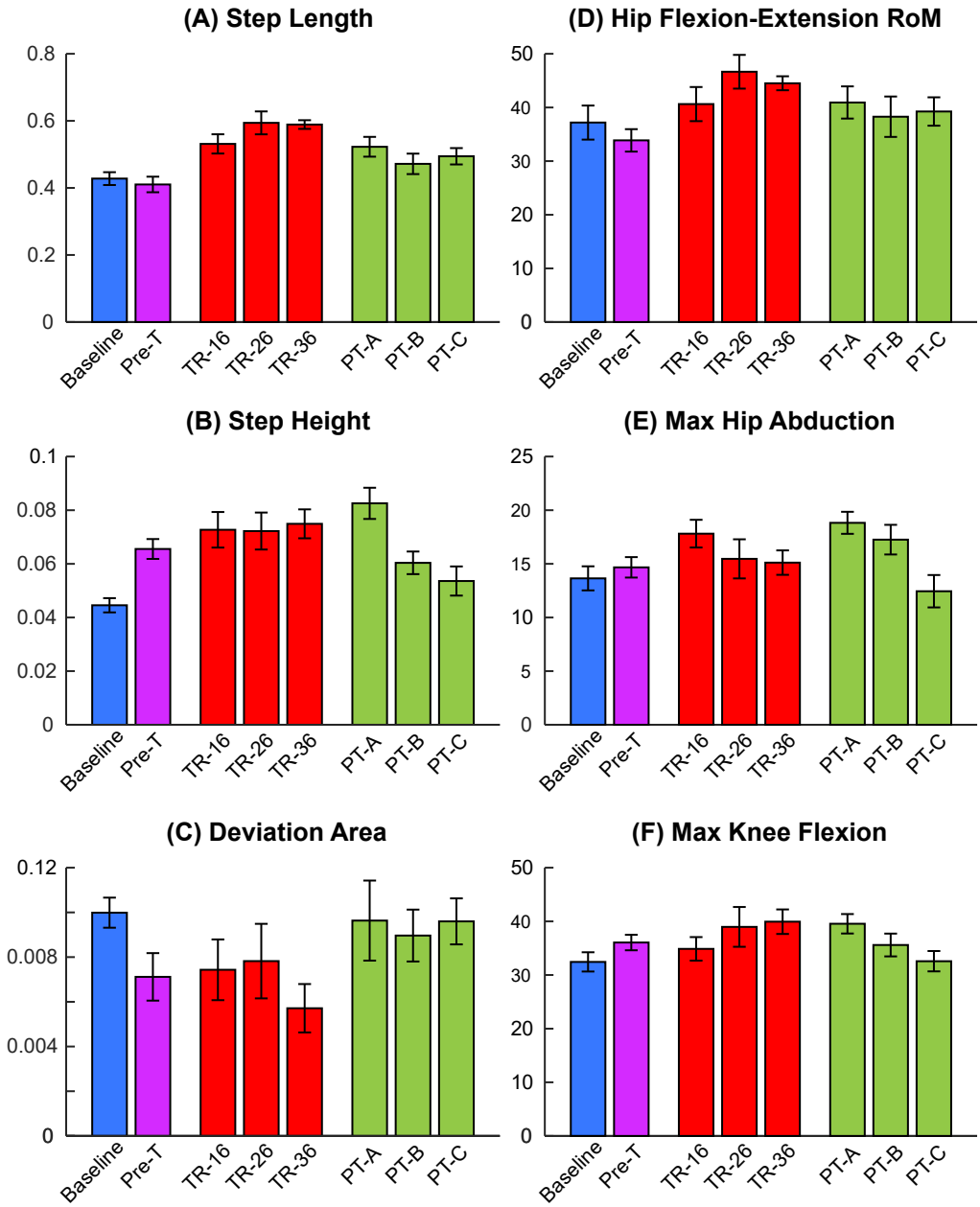


Figure 5.5: Gait Parameters of the Stroke Patient. The gait parameters are averaged during an one-minute portion of each session, with the mean and standard deviation shown in the figure.

sessions, the step height and length both reduced compared to PT-A sessions, but were still larger than the baseline session.

Overall, the results showed that the gait of the patient improved during the training

session with C-ALEX. The stroke patient was able to retain some of the changes to the post-training sessions after taking off C-ALEX.

5.4 Discussion

The Pre-T session showed the changes in gait of the stroke patient while wearing C-ALEX in the transparent mode. The gait of the patient was close to the baseline gait during the free-walking session, except for an increase in step height. After removing C-ALEX during the PT-B session, a decrease in step height was also observed. This result is different from the experiment with healthy subjects, in which wearing C-ALEX caused an increase in step length but almost no changes in step height. Whether this change in step height is unique to this particular patient requires further studies.

After adding the force assistance from C-ALEX during the training sessions, there is an increase in step length and step height. The deviation area reduced as the gait of the patient became closer to the target gait. These changes suggest that the force assistance from C-ALEX combined with the visual feedback can help the stroke patient follow the target gait. In the joint space, the hip flexion-extension RoM increased during the training compared with baseline gait. In particular, during TR-26, the hip flexion-extension RoM increased by 25.1%, very close to target gait which has a 25% increase in hip flexion-extension RoM. The increase in knee flexion is below the 50% target in all TR sessions. It is possible that the 50% increase in knee flexion is too difficult for the subject.

After the training, the subject partially retained the increase in step height and step length compared to the baseline, even after C-ALEX was removed. The deviation area

was not as small as the training sessions, but being able to follow the exact target ankle trajectory was not as important as taking a longer and higher step for the stroke patient. Therefore, it is fair to state that the training with C-ALEX brought positive changes to the gait of the stroke patient.

During the experiment, the changes in step height were accompanied by both an increase in knee flexion, and an increase in the hip abduction. While the increase in knee flexion is desirable, the increase in hip abduction is not. One limitation of the current force field controller is that the force assistance is based on the ankle position. The increase in hip abduction can increase step height and bringing the ankle closer to the target trajectory, therefore causing C-ALEX to reduce the assistance level. One way to alleviate this issue is to have the controller of C-ALEX providing assistance based on the joint angle instead of the ankle position. This approach will be tested in future experiments.

5.5 Conclusion

Overall, the experiment showed promising results on a single stroke patient. The patient was able to increase the step length and step height during the training, and this increase got partly carried over in the post-training sessions after the removing C-ALEX. This result showed that C-ALEX could be potentially used as a training device in stroke rehabilitation. Future experiments should test C-ALEX on additional stroke patients to validate the findings of this study, and also investigate multiple sessions training with C-ALEX and long-term retention.

Contributions and Future Works

Contributions

This thesis presented the motivation, prototype design, and experimental result of a novel leg exoskeleton named C-ALEX, which employed a unique cable-driven design that eliminated the rigid-linked structure commonly found in exoskeletons. While there exist cable-driven exoskeletons in the literature, they are either limited to a single joint design or partially use rigid-linked structures. C-ALEX is the first cable-driven exoskeleton that can provide multiple joints assistance to the user without using any rigid-link structure.

C-ALEX demonstrated the two main benefits of the cable-driven design. First, C-ALEX is exceptionally lightweight. Combining the cable-driven actuation with 3D printed cuffs makes the moving weight of C-ALEX to be merely 1.6 kg on one leg, and a total weight of 4.3 kg on the body of the user. Extending C-ALEX into a bilateral design will only add 2.5 kg and bring the total weight to 6.8 kg. This weight is much lower than rigid-linked exoskeletons, which typically add more than 10 kg of weight on the human user. The second advantage is that without any mechanical joints, C-ALEX does not restrict the motion of the user. The mechanical joints of the rigid-linked exoskeleton bring multiple issues: the number of joints may not be enough to allow all DoFs of the user, and the

joints may not be aligned with the human joint. By not having any mechanical joints, C-ALEX avoided these issues. The issue of joint misalignment is no longer a concern, and the user can move naturally without any constraint.

Besides contributions in the mechanical design of exoskeletons, Chapter 2 of this thesis also presented a novel approach to control the torque applied to the user using cable tensions. Finding the desired cable tension to generate the desired joint torque requires solving a set of linear equations under constraint, which may have no solution at certain configurations. This leads to discontinuities and abrupt changes in cable tension, which will cause jerky motion to the user. The new method solves this issue by finding an approximate solution instead of an exact one, thus guaranteeing a solution can always be found. This method ensures continuous cable tension output and reduced the jerkiness in the motion of the user.

Another contribution of this thesis is the assist-as-needed controller that estimates the current phase of gait based on the past ankle trajectory. Gait phase estimation usually requires additional sensors placed in the shoe of the user, which adds extra complexity to the system. The gait phase estimation algorithm detailed in Chapter 2 only relies on the joint angle of the leg, therefore does not require additional sensors. The algorithm is simple and efficient, can provide good accuracy and quickly adapts to the change in the gait of the user. Gait phase estimation is widely used in both training and assistive leg exoskeletons, therefore the proposed algorithm has the potential to be adopted in other applications.

The lightweight nature of C-ALEX makes it a good platform to study the effect of exoskeleton weight and inertia on the gait of the user. By attaching extra masses to

C-ALEX and manipulating weight compensation in its controller, C-ALEX can simulate exoskeletons with different weight and inertia. Chapter 3 of this thesis includes a study that explored the effect of weight and inertia on the natural gait of the user, and showed that the knee flexion was predominantly affected by added weight, while the changes in step length were almost exclusively due to added inertia.

C-ALEX is the first leg exoskeleton for gait rehabilitation using a cable-driven design, and this thesis proved its effectiveness in gait training. Chapter 4 of this thesis showed that healthy individuals can adapt to a new gait after training with C-ALEX. The effectiveness of C-ALEX is comparable with traditional leg exoskeletons based on rigid links. Chapter 5 tested C-ALEX on a stroke patient and showed improvements in several gait parameters, such as step length and height, after 40 minutes training.

Future Works

The current work showed the design and evaluation of a cable-driven leg exoskeleton C-ALEX for gait training with an assist-as-needed controller. There are many aspects of C-ALEX that can be improved to enhance its functionality.

On the aspect of mechanical design, the current 3D printed structure of C-ALEX is designed conservatively to avoid part failure. No part showed signs of failure after extensive experiments, which verifies the robustness of the current design, but also suggests that the parts can be redesigned to be lighter and more compact to further reduce the weight and inertia of C-ALEX. Another advantage of 3D printing is that the parts of C-ALEX can be potentially customized to each individual user to improve its fit.

Another aspect that can be improved with C-ALEX is to replace the Vicon motion capture system. Vicon system requires a large working space and is expensive to acquire. One possible alternative is to use inertial measurement unit (IMU) sensors on each segment of the leg to measure the joint angles. However, IMUs are susceptible to drift and need to be recalibrated periodically. Recent development in VR technologies offers another option, which is to fuse IMU sensors with optical sensors to obtain reliable orientation measurements without drift. Currently, VIVE Tracker [1] sensors are being tested to study its accuracy and its potential application on C-ALEX. It offers a tracking solution in a small profile at low cost.

In terms of the control and gait training experimentation, C-ALEX is not limited to the assist-as-needed approach presented in this thesis. Given the joint torque control capability of C-ALEX, other kinds of training paradigms such as error augmentation [79] and viscosity field [72] can also be implemented on C-ALEX. Because of the transparent nature of C-ALEX, the user will receive more assistance force and less disturbance from the weight and inertia of the exoskeleton. Therefore, future experiments can explore other control approaches for C-ALEX and compare the training effect of C-ALEX with other exoskeletons.

Besides improving the design and functionality of C-ALEX, another important future direction of C-ALEX is to bring it to over-ground gait-training and investigate an important scientific question: is gait training over-ground better than training on the treadmill? Different studies have compared over-ground walking with treadmill walking in the healthy population, and observed differences in the leg kinetics, muscle EMG,

[1] Valve, WA, US

and metabolic cost [80]–[82]. In terms of gait training, a recent review has compared robotics-assisted gait training with traditional over-ground gait training on spinal cord injury patients, and the results favor the traditional over-ground training [83]. It is not clear whether the difference comes from the use of the robotic device or the two different walking conditions. One way to answer this question is to compare over-ground training with treadmill training using the same exoskeleton. C-ALEX has the potential to be transformed into a mobile exoskeleton that can be used either over-ground or on a treadmill. To become a mobile exoskeleton, the motors and electronics of C-ALEX need to be integrated into a compact package that can be placed either on the body of the user or on a mobile cart that follows the user. The Vicon system used in the current version of C-ALEX also needs to be replaced by on-body sensors such as IMUs for mobile use. After these transformations, C-ALEX would be able to be used for both over-ground gait training and treadmill-based gait training. Given the transparent nature of C-ALEX, it would become an ideal platform to investigate the differences between the two types of gait training approaches, possibly settle the long-time debate of which training approach is superior.

Bibliography

[1] K. N. Gregorczyk, L. Hasselquist, J. M. Schiffman, C. K. Bensel, J. P. Obusek, and D. J. Gutekunst, "Effects of a lower-body exoskeleton device on metabolic cost and gait biomechanics during load carriage," *Ergonomics*, vol. 53, no. 10, pp. 1263–1275, Oct. 2010.

[2] S. Jacobsen, M. Olivier, F. Smith, D. Knutti, R. Johnson, G. Colvin, and W. Scroggin, "Research Robots for Applications in Artificial Intelligence, Teleoperation and Entertainment," *The International Journal of Robotics Research*, vol. 23, no. 4-5, pp. 319–330, Apr. 2004.

[3] K. A. Strausser and H. Kazerooni, "The development and testing of a human machine interface for a mobile medical exoskeleton," in *2011 IEEE/RSJ International Conference on Intelligent Robots and Systems*, Sep. 2011, pp. 4911–4916.

[4] M. Talaty, A. Esquenazi, and J. E. Briceño, "Differentiating ability in users of the ReWalkTM powered exoskeleton: An analysis of walking kinematics," in *2013 IEEE 13th International Conference on Rehabilitation Robotics (ICORR)*, Jun. 2013, pp. 1–5.

[5] C. Krishnan, R. Ranganathan, S. S. Kantak, Y. Y. Dhaher, and W. Z. Rymer, "Active robotic training improves locomotor function in a stroke survivor," *Journal of NeuroEngineering and Rehabilitation*, vol. 9, p. 57, 2012.

[6] S. Banala, S. H. Kim, S. Agrawal, and J. Scholz, "Robot Assisted Gait Training With Active Leg Exoskeleton (ALEX)," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 17, no. 1, pp. 2–8, Feb. 2009.

[7] A. Pennycott, D. Wyss, H. Vallery, V. Klamroth-Marganska, and R. Riener, "Towards more effective robotic gait training for stroke rehabilitation: A review," *Journal of NeuroEngineering and Rehabilitation*, vol. 9, p. 65, Sep. 2012.

[8] S. Maggioni, A. Melendez-Calderon, E. van Asseldonk, V. Klamroth-Marganska, L. Lünenburger, R. Riener, and H. van der Kooij, "Robot-aided assessment of lower extremity functions: A review," *Journal of NeuroEngineering and Rehabilitation*, vol. 13, p. 72, 2016.

[9] J.-M. Belda-Lois, S. Mena-del Horno, I. Bermejo-Bosch, J. C. Moreno, J. L. Pons, D. Farina, M. Iosa, M. Molinari, F. Tamburella, A. Ramos, A. Caria, T. Solis-Escalante, C. Brunner, and M. Rea, "Rehabilitation of gait after stroke: A review towards a top-down approach," *Journal of NeuroEngineering and Rehabilitation*, vol. 8, p. 66, 2011.

[10] E. J. Benjamin, M. J. Blaha, S. E. Chiuve, M. Cushman, S. R. Das, R. Deo, S. D. d. Ferranti, J. Floyd, M. Fornage, C. Gillespie, C. R. Isasi, M. C. Jiménez, L. C. Jordan, S. E. Judd, D. Lackland, J. H. Lichtman, L. Lisabeth, S. Liu, C. T. Longenecker, R. H. Mackey, K. Matsushita, D. Mozaffarian, M. E. Mussolino, K. Nasir, R. W. Neumar, L. Palaniappan, D. K. Pandey, R. R. Thiagarajan, M. J. Reeves, M. Ritchey, C. J. Rodriguez, G. A. Roth, W. D. Rosamond, C. Sasson, A. Towfighi, C. W. Tsao, M. B. Turner, S. S. Virani, J. H. Voeks, J. Z. Willey, J. T. Wilkins, J. H. Wu, H. M. Alger, S. S. Wong, P. Muntner, and O. b. o. t. A. H. A. S. C. a. S. S. Subcommittee, "Heart Disease and Stroke Statistics—2017 Update: A Report From the American Heart Association," *Circulation*, CIR.0000000000000485, Jan. 2017.

[11] U. Centers for Disease Control and Prevention, "Prevalence and most common causes of disability among adults–United States, 2005," *MMWR: Morbidity and mortality weekly report*, vol. 58, no. 16, pp. 421–426, 2009.

[12] G. Morone, S. Paolucci, A. Cherubini, D. De Angelis, V. Venturiero, P. Coiro, and M. Iosa, "Robot-assisted gait training for stroke patients: Current state of the art and perspectives of robotics," *Neuropsychiatric Disease and Treatment*, vol. 13, pp. 1303–1311, May 2017.

[13] R. Riener, L&#xfc, L. Nenburger, I. C. Maier, G. Colombo, and V. Dietz, "Locomotor Training in Subjects with Sensori-Motor Deficits: An Overview of the Robotic Gait Orthosis Lokomat," *Journal of Healthcare Engineering*, vol. 1, no. 2, pp. 197–216, 2010.

[14] A. Mayr, M. Kofler, E. Quirbach, H. Matzak, K. Fröhlich, and L. Saltuari, "Prospective, Blinded, Randomized Crossover Study of Gait Rehabilitation in Stroke Patients Using the Lokomat Gait Orthosis," *Neurorehabilitation and Neural Repair*, vol. 21, no. 4, pp. 307–314, Jul. 2007.

[15] J. Hidler, D. Nichols, M. Pelliccio, K. Brady, D. D. Campbell, J. H. Kahn, and T. G. Hornby, "Multicenter Randomized Clinical Trial Evaluating the Effectiveness of the Lokomat in Subacute Stroke," *Neurorehabilitation and Neural Repair*, vol. 23, no. 1, pp. 5–13, Jan. 2009.

[16] S. Fisher, L. Lucas, and T. A. Thrasher, "Robot-Assisted Gait Training for Patients with Hemiparesis Due to Stroke," *Topics in Stroke Rehabilitation*, vol. 18, no. 3, pp. 269–276, May 2011.

[17] S. K. Banala, S. K. Agrawal, and J. P. Scholz, "Active Leg Exoskeleton (ALEX) for Gait Rehabilitation of Motor-Impaired Patients," in *2007 IEEE 10th International Conference on Rehabilitation Robotics*, Jun. 2007, pp. 401–407.

[18] K. Winfree, P. Stegall, and S. Agrawal, "Design of a minimally constraining, passively supported gait training exoskeleton: ALEX II," in *2011 IEEE International Conference on Rehabilitation Robotics (ICORR)*, Jun. 2011, pp. 1–6.

[19] D. Zanotto, P. Stegall, and S. Agrawal, "ALEX III: A novel robotic platform with 12 DOFs for human gait training," in *2013 IEEE International Conference on Robotics and Automation (ICRA)*, May 2013, pp. 3914–3919.

[20] S. Srivastava, P. C. Kao, S. H. Kim, P. Stegall, D. Zanotto, J. S. Higginson, S. K. Agrawal, and J. P. Scholz, "Assist-as-Needed Robot-Aided Gait Training Improves Walking Function in Individuals Following Stroke," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 23, no. 6, pp. 956–963, Nov. 2015.

[21] R. J. Farris, H. A. Quintero, and M. Goldfarb, "Preliminary Evaluation of a Powered Lower Limb Orthosis to Aid Walking in Paraplegic Individuals," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 19, no. 6, pp. 652–659, Dec. 2011.

[22] M. Bortole, A. Venkatakrishnan, F. Zhu, J. C. Moreno, G. E. Francisco, J. L. Pons, and J. L. Contreras-Vidal, "The h2 robotic exoskeleton for gait rehabilitation after stroke: Early findings from a clinical study," *Journal of NeuroEngineering and Rehabilitation*, vol. 12, p. 54, Jun. 2015.

[23] *Mobility Training Using Exoskeletons for Functional Recovery After Stroke.* [Online]. Available: `https://clinicaltrials.gov/ct2/show/NCT02128152`.

[24] H. Watanabe, N. Tanaka, T. Inuta, H. Saitou, and H. Yanagi, "Locomotion Improvement Using a Hybrid Assistive Limb in Recovery Phase Stroke Patients: A Randomized Controlled Pilot Study," *Archives of Physical Medicine and Rehabilitation*, vol. 95, no. 11, pp. 2006–2012, Nov. 2014.

[25] H. Fukuda, K. Samura, O. Hamada, K. Saita, T. Ogata, E. Shiota, Y. Sankai, and T. Inoue, "Effectiveness of Acute Phase Hybrid Assistive Limb Rehabilitation in Stroke Patients Classified by Paralysis Severity," *Neurologia medico-chirurgica*, vol. 55, no. 6, pp. 487–492, 2015.

[26] D. R. Louie and J. J. Eng, "Powered robotic exoskeletons in post-stroke rehabilitation of gait: A scoping review," *Journal of NeuroEngineering and Rehabilitation*, vol. 13, p. 53, Jun. 2016.

[27] G. Colombo, M. Joerg, R. Schreier, and V. Dietz, "Treadmill training of paraplegic patients using a robotic orthosis," *Journal of Rehabilitation Research and Development; Washington*, vol. 37, no. 6, pp. 693–700, Dec. 2000.

[28] K. van Kammen, A. M. Boonstra, L. H. V. van der Woude, H. A. Reinders-Messelink, and R. den Otter, "Differences in muscle activity and temporal step parameters between Lokomat guided walking and treadmill walking in post-stroke hemiparetic patients and healthy walkers," *Journal of NeuroEngineering and Rehabilitation*, vol. 14, p. 32, 2017.

[29] A. Kaelin-Lang, L. Sawaki, and L. G. Cohen, "Role of Voluntary Drive in Encoding an Elementary Motor Memory," *Journal of Neurophysiology*, vol. 93, no. 2, pp. 1099–1103, Feb. 2005.

[30] N. Hogan, "Impedance Control: An Approach to Manipulation: Part II—Implementation," *Journal of Dynamic Systems, Measurement, and Control*, vol. 107, no. 1, pp. 8–16, Mar. 1985.

[31] L. L. Cai, A. J. Fong, C. K. Otoshi, Y. Liang, J. W. Burdick, R. R. Roy, and V. R. Edgerton, "Implications of Assist-As-Needed Robotic Step Training after a Complete Spinal Cord Injury on Intrinsic Strategies of Motor Learning," *Journal of Neuroscience*, vol. 26, no. 41, pp. 10 564–10 568, Oct. 2006.

[32] A. Duschau-Wicke, J. v. Zitzewitz, A. Caprez, L. Lunenburger, and R. Riener, "Path Control: A Method for Patient-Cooperative Robot-Aided Gait Rehabilitation," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 18, no. 1, pp. 38–48, Feb. 2010.

[33] D. Aoyagi, W. E. Ichinose, S. J. Harkema, D. J. Reinkensmeyer, and J. E. Bobrow, "A Robot and Control Algorithm That Can Synchronously Assist in Naturalistic Motion During Body-Weight-Supported Gait Training Following Neurologic Injury," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 15, no. 3, pp. 387–400, Sep. 2007.

[34] J. Veneman, R. Kruidhof, E. E. G. Hekman, R. Ekkelenkamp, E. H. F. Van Asseldonk, and H. van der Kooij, "Design and Evaluation of the LOPES Exoskeleton Robot for Interactive Gait Rehabilitation," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 15, no. 3, pp. 379–386, Sep. 2007.

[35] A. Duschau-Wicke, A. Caprez, and R. Riener, "Patient-cooperative control increases active participation of individuals with SCI during robot-aided gait training," *Journal of NeuroEngineering and Rehabilitation*, vol. 7, p. 43, 2010.

[36] X. Jin, Y. Cai, A. Prado, and S. K. Agrawal, "Effects of exoskeleton weight and inertia on human walking," in *2017 IEEE International Conference on Robotics and Automation (ICRA)*, May 2017, pp. 1772–1777.

[37] D. A. Winter, *Biomechanics and Motor Control of Human Movement*. John Wiley & Sons, Oct. 2009, Google-Books-ID: _bFHL08IWfwC.

[38] M. Cenciarini and A. Dollar, "Biomechanical considerations in the design of lower limb exoskeletons," in *2011 IEEE International Conference on Rehabilitation Robotics (ICORR)*, Jun. 2011, pp. 1–6.

[39] P. Stegall, K. Winfree, D. Zanotto, and S. Agrawal, "Rehabilitation Exoskeleton Design: Exploring the Effect of the Anterior Lunge Degree of Freedom," *IEEE Transactions on Robotics*, vol. 29, no. 4, pp. 838–846, Aug. 2013.

[40] D. Zanotto, T. Lenzi, P. Stegall, and S. Agrawal, "Improving transparency of powered exoskeletons using force/torque sensors on the supporting cuffs," in *2013 IEEE International Conference on Rehabilitation Robotics (ICORR)*, Jun. 2013, pp. 1–6.

[41] E. Colgate and N. Hogan, "An analysis of contact instability in terms of passive physical equivalents," in *1989 International Conference on Robotics and Automation Proceedings*, May 1989, 404–409 vol.1.

[42] W. v. Dijk, H. v. d. Kooij, B. Koopman, E. H. F. v. Asseldonk, and H. v. d. Kooij, "Improving the transparency of a rehabilitation robot by exploiting the cyclic behaviour of walking," in *2013 IEEE International Conference on Rehabilitation Robotics (ICORR)*, Jun. 2013, pp. 1–8.

[43] J. Hidler, W. Wisman, and N. Neckel, "Kinematic trajectories while walking within the Lokomat robotic gait-orthosis," *Clinical Biomechanics*, vol. 23, no. 10, pp. 1251–1259, Dec. 2008.

[44] J. F. Veneman, J. Menger, E. H. F. van Asseldonk, F. C. T. van der Helm, and H. van der Kooij, "Fixating the pelvis in the horizontal plane affects gait characteristics," *Gait & Posture*, vol. 28, no. 1, pp. 157–163, Jul. 2008.

[45] J. M. Hidler and A. E. Wall, "Alterations in muscle activation patterns during robotic-assisted walking," *Clinical Biomechanics*, vol. 20, no. 2, pp. 184–193, Feb. 2005.

[46] J. Meuleman, E. v. Asseldonk, G. v. Oort, H. Rietman, and H. v. d. Kooij, "LOPES II: Design and Evaluation of an Admittance Controlled Gait Training Robot With Shadow-Leg Approach," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 24, no. 3, pp. 352–363, Mar. 2016.

[47] Y. Mao, X. Jin, G. G. Dutta, J. P. Scholz, and S. K. Agrawal, "Human Movement Training With a Cable Driven ARm EXoskeleton (CAREX)," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 23, no. 1, pp. 84–92, Jan. 2015, WOS:000347913400009.

[48] J. Zhang, P. Fiers, K. A. Witte, R. W. Jackson, K. L. Poggensee, C. G. Atkeson, and S. H. Collins, "Human-in-the-loop optimization of exoskeleton assistance during walking," *Science*, vol. 356, no. 6344, pp. 1280–1284, Jun. 2017.

[49] A. Alamdari and V. Krovi, "Design and Analysis of a Cable-Driven Articulated Rehabilitation System for Gait Training," *Journal of Mechanisms and Robotics*, vol. 8, no. 5, pp. 051 018–051 018, May 2016.

[50] M. Wu, T. G. Hornby, J. M. Landry, H. Roth, and B. D. Schmit, "A cable-driven locomotor training system for restoration of gait in human SCI," *Gait & Posture*, vol. 33, no. 2, pp. 256–260, Feb. 2011.

[51] K. Schmidt and R. Riener, "MAXX: Mobility Assisting teXtile eXoskeleton that Exploits Neural Control Synergies," in *Converging Clinical and Engineering Research on Neurorehabilitation II*, ser. Biosystems & Biorobotics, DOI: 10.1007/978-3-319-46669-9_89, Springer, Cham, 2017, pp. 539–543.

[52] B. T. Quinlivan, S. Lee, P. Malcolm, D. M. Rossi, M. Grimmer, C. Siviy, N. Karavas, D. Wagner, A. Asbeck, I. Galiana, and C. J. Walsh, "Assistance magnitude versus metabolic cost reductions for a tethered multiarticular soft exosuit," *Science Robotics*, vol. 2, no. 2, eaah4416, Jan. 2017.

[53] A. Asbeck, R. Dyer, A. Larusson, and C. Walsh, "Biologically-inspired soft exosuit," in *2013 IEEE International Conference on Rehabilitation Robotics (ICORR)*, Jun. 2013, pp. 1–8.

[54] A. T. Asbeck, K. Schmidt, I. Galiana, D. Wagner, and C. J. Walsh, "Multi-joint soft exosuit for gait assistance," in *2015 IEEE International Conference on Robotics and Automation (ICRA)*, May 2015, pp. 6197–6204.

[55] S. Lee, S. Crea, P. Malcolm, I. Galiana, A. Asbeck, and C. Walsh, "Controlling negative and positive power at the ankle with a soft exosuit," in *2016 IEEE International Conference on Robotics and Automation (ICRA)*, May 2016, pp. 3509–3515.

[56] A. T. Asbeck, S. M. De Rossi, K. G. Holt, and C. J. Walsh, "A biologically inspired soft exosuit for walking assistance," *The International Journal of Robotics Research*, vol. 34, no. 6, pp. 744–762, May 2015.

[57] J. Bae, S. M. M. D. Rossi, K. O'Donnell, K. L. Hendron, L. N. Awad, T. R. T. D. Santos, V. L. D. Araujo, Y. Ding, K. G. Holt, T. D. Ellis, and C. J. Walsh, "A soft exosuit for

patients with stroke: Feasibility study with a mobile off-board actuation unit," in *2015 IEEE International Conference on Rehabilitation Robotics (ICORR)*, Aug. 2015, pp. 131–138.

[58] S. H. Collins, M. B. Wiggin, and G. S. Sawicki, "Reducing the energy cost of human walking using an unpowered exoskeleton," *Nature*, vol. 522, no. 7555, pp. 212–215, Jun. 2015.

[59] A. J. Young and D. P. Ferris, "State of the Art and Future Directions for Lower Limb Robotic Exoskeletons," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 25, no. 2, pp. 171–182, Feb. 2017.

[60] G. Chen, C. K. Chan, Z. Guo, and H. Yu, "A Review of Lower Extremity Assistive Robotic Exoskeletons in Rehabilitation Therapy," *Critical Reviews™ in Biomedical Engineering*, vol. 41, no. 4-5, 2013.

[61] D. Zanotto, P. Stegall, and S. K. Agrawal, "Adaptive assist-as-needed controller to improve gait symmetry in robot-assisted gait training," in *2014 IEEE International Conference on Robotics and Automation (ICRA)*, May 2014, pp. 724–729.

[62] E. H. F. v. Asseldonk, J. F. Veneman, R. Ekkelenkamp, J. H. Buurke, F. C. T. v. d. Helm, and H. v. d. Kooij, "The Effects on Kinematics and Muscle Activity of Walking in a Robotic Gait Trainer During Zero-Force Control," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 16, no. 4, pp. 360–370, Aug. 2008.

[63] H. Vallery, A. Duschau-Wicke, and R. Riener, "Optimized passive dynamics improve transparency of haptic devices," in *IEEE International Conference on Robotics and Automation, 2009. ICRA '09*, May 2009, pp. 301–306.

[64] R. C. Browning, J. R. Modica, R. Kram, and A. Goswami, "The Effects of Adding Mass to the Legs on the Energetics and Biomechanics of Walking," *Medicine & Science in Sports & Exercise*, vol. 39, no. 3, pp. 515–525, Mar. 2007.

[65] T. D. Royer and P. E. Martin, "Manipulations of Leg Mass and Moment of Inertia: Effects on Energy Cost of Walking," *Medicine & Science in Sports & Exercise*, vol. 37, no. 4, pp. 649–656, Apr. 2005.

[66] K. G. Holt, J. Hamill, and R. O. Andres, "The force-driven harmonic oscillator as a model for human locomotion," *Human Movement Science*, vol. 9, no. 1, pp. 55–68, Feb. 1990.

[67] J. W. Noble and S. D. Prentice, "Adaptation to unilateral change in lower limb mechanical properties during human walking," *Experimental Brain Research*, vol. 169, no. 4, pp. 482–495, Mar. 2006.

[68] J. H. Meuleman, E. H. van Asseldonk, and H. van der Kooij, "The effect of directional inertias added to pelvis and ankle on gait," *Journal of NeuroEngineering and Rehabilitation*, vol. 10, p. 40, 2013.

[69] L. Wolfson, "Gait and Balance Dysfunction: A Model of the Interaction of Age and Disease," *The Neuroscientist*, vol. 7, no. 2, pp. 178–183, Apr. 2001.

[70] *Funny Cats Compilation (Most Popular) Part 1.* [Online]. Available: `https://youtu.be/3EIbWjkimAs`.

[71] S. Banala, S. Agrawal, S. H. Kim, and J. Scholz, "Novel Gait Adaptation and Neuromotor Training Results Using an Active Leg Exoskeleton," *IEEE/ASME Transactions on Mechatronics*, vol. 15, no. 2, pp. 216–225, Apr. 2010.

[72] P. Stegall, D. Zanotto, and S. K. Agrawal, "Variable Damping Force Tunnel for Gait Training Using ALEX III," *IEEE Robotics and Automation Letters*, vol. 2, no. 3, pp. 1495–1501, Jul. 2017.

[73] R. Ranganathan, C. Krishnan, Y. Y. Dhaher, and W. Z. Rymer, "Learning new gait patterns: Exploratory muscle activity during motor learning is not predicted by motor modules," *Journal of Biomechanics*, vol. 49, no. 5, pp. 718–725, Mar. 2016.

[74] J. L. Emken, J. E. Bobrow, and D. J. Reinkensmeyer, "Robotic movement training as an optimization problem: Designing a controller that assists only as needed," in *9th International Conference on Rehabilitation Robotics, 2005. ICORR 2005.*, Jun. 2005, pp. 307–312.

[75] X. Jin, X. Cui, and S. K. Agrawal, "Design of a cable-driven active leg exoskeleton (c-ALEX) and gait training experiments with human subjects," in *2015 IEEE International Conference on Robotics and Automation (ICRA)*, May 2015, pp. 5578–5583.

[76] D. J. Clark, "Automaticity of walking: Functional significance, mechanisms, measurement and rehabilitation strategies," *Frontiers in Human Neuroscience*, vol. 9, May 2015.

[77] O. Beauchet, V. Dubost, F. R. Herrmann, and R. W. Kressig, "Stride-to-stride variability while backward counting among healthy young adults," *Journal of NeuroEngineering and Rehabilitation*, vol. 2, p. 26, Aug. 2005.

[78] J. F. Israel, D. D. Campbell, J. H. Kahn, and T. G. Hornby, "Metabolic Costs and Muscle Activity Patterns During Robotic- and Therapist-Assisted Treadmill Walking in Individuals With Incomplete Spinal Cord Injury," *Physical Therapy*, vol. 86, no. 11, pp. 1466–1478, Nov. 2006.

[79] M.-H. Milot, L. Marchal-Crespo, C. S. Green, S. C. Cramer, and D. J. Reinkensmeyer, "Comparison of error-amplification and haptic-guidance training techniques for learning of a timing-based motor task by healthy individuals," *Experimental Brain Research*, vol. 201, no. 2, pp. 119–131, Mar. 2010.

[80] G. J. V. I. Schenau, "Some fundamental aspects of the biomechanics of overground versus treadmill locomotion," *Medicine & Science in Sports & Exercise*, vol. 12, no. 4, pp. 257–261, Jan. 1980.

[81] S. J. Lee and J. Hidler, "Biomechanics of overground vs. treadmill walking in healthy individuals," *Journal of Applied Physiology*, vol. 104, no. 3, pp. 747–755, Mar. 2008.

[82] J.-P. Martin and Q. Li, "Overground vs. treadmill walking on biomechanical energy harvesting: An energetics and EMG study," *Gait & Posture*, vol. 52, pp. 124–128, Feb. 2017.

[83] J. Mehrholz, L. A. Harvey, S. Thomas, and B. Elsner, "Is body-weight-supported treadmill training or robotic-assisted gait training superior to overground gait training and other forms of physiotherapy in people with spinal cord injury? a systematic review," *Spinal Cord*, vol. 55, no. 8, pp. 722–729, Aug. 2017.