Building Better Exoskeletons:
Understanding How Design Affects Robot Assisted Gait Training

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Abstract

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Physical therapy is a field with ever increasing demands as the population ages, resulting in a larger number of individuals living with impairments. Therapy is both physically intensive and time intensive for physical therapists, and can require more than one therapist per patient. The use of technology can reduce both these physical and time demands if appropriately applied, while improving repeatability and providing quantitative evaluation of performance. Through these abilities, it may also improve the quality of life for patients. The work presented here explores how the mechanical and controller design of exoskeletons can be used to improve adaptations to new gait patterns in healthy individuals. Armed with this knowledge, new treatment methods can be adapted, applied, and validated for impaired populations with the intention of recovering a more natural gait pattern.

First, the ALEX II device is presented. It is a unilateral device, designed to aid in gait training for stroke survivors. The previous version, ALEX I, had several limitations in terms of pelvic freedom, leg range of motion, and the support of the gravitational load. ALEX II was designed to address these issues. Next, a study is presented, using healthy young adults (N=30), in which ALEX II was used to explore how the amount of freedom allowed at the pelvis during gait training affects the level of adaptation subjects are able to achieve. This was evaluated for five separate configurations which resemble existing exoskeletons. It was found that intermediate levels of pelvic freedom degrade the amount of adaptation and that pelvic translation contributes more to this effect than hip abduction/adduction.

The next work concerns the design of ALEX III, a bilateral device with twelve active degrees-of-freedom. ALEX III was created to increase the ability to explore the functionality required for gait training, which is why it is capable of controlling 4 degrees-of-freedom at each leg, and 4 degrees-of-freedom at the pelvis. This is followed by the the design of a new type of haptic feedback which utilizes a variable, viscous damping field, which increases the damping coeffiecent as the subject moves away from a specified path.

This feedback type was tested in a set of experiments in healthy young adults. The first study (N=32) compared four different settings for the new feedback, finding that while all groups demonstrated adaptations in gait, the lowest rate of change of the damping field exhibited less adaptation. The final study (N=36) compared this haptic feedback to two previously used haptic feedback types. The previously used feedback strategies used a force that pushed the leg either towards or away from the desired path. All three of these strategies were found to produce similar levels of adaptation, however the damping field used much less external force. These findings may change the way exoskeletons for gait training are designed and increase their accessibility.

While all the findings need to be validated in impaired populations they can still inform the design of future exoskeletons. The first finding, that providing an intermediate amount of freedom to the pelvis can interfere with gait training, suggests that future devices should have very high amounts of freedom or very restricted pelvic motions. The final finding, that damping fields can be used to induce gait adaptations using a much lower force, can drastically change exoskeleton design and how robotic therapy is provided. Exoskeletons can be made lighter as a result of the force being highly reduced so that lighter weight components can be used, and the dissipative nature of the force reduces dependence on heavy power sources because regenerative breaking can be used to power the device. These factors also make it possible to for devices to be used overground, which may make training more transferable to the real world.

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Chapter 1

Introduction

1.1 Motivation

In United States alone, there are 795,000 strokes each year. More than 600,000 of these occur in individuals who have not previously had a stroke [1]. This equates to roughly one stroke every 40 seconds. The survival rate for first time stroke victims is over 50%. Since most of these survivors will require some form of rehabilitation to regain their functional abilities, this creates an increasing demand on the physical therapy community.

There are many approaches currently employed for gait oriented physical therapy. Strengthening, rhythmic auditory stimulation, treadmill training, and overground training will be examined in this section but this is by no means an exhaustive list of treatment methodologies currently employed. First, strengthening exercises for gait rehabilitation can produce some advantages in terms of increased ability to generate force, or reduced risk of osteoporosis, but several meta-analysis studies have found that they do not produce improvements in walking speed [2]. As a result of sensory and muscle coordination deficits, practicing the specific task is now thought to be a crucial element in producing improvements in that task.

Task specific training focuses on actively performing a targeted task. For gait training this would consist of walking. This is typically broken into two categories, overground and treadmill training; each with its own advantages and disadvantages. Overground training provides training most similar to what would be encountered in daily life, and challenges such as turning and navigating obstacles can easily be added.

However, it is more challenging to add assistance and it requires significant floorspace. While body weight support (BWS) can be provided overground, it requires the installation of an overhead track in the clinic which limits a patient to walking under the track, or uses a rolling frame which therapists must push with the patient. Using a treadmill instead provides a less natural environment as the speed is fixed and there is no turning. However, using a treadmill provides the ability to perform a large number of repetitions in a short period of time, and it makes it easier to provide aid as therapist can sit beside the treadmill and manually move the legs to help perform more typical movement patterns. Adding BWS to treadmill training is relatively easy by leaving the same rolling frame stationary over the treadmill or using a treadmill with an overhead harness.

Manually assisted therapy can require up to three physical therapist, one moving each leg and one moving the pelvis. This is expensive as it requires the time of three therapists, and it is physically demanding on the physical therapist [3] to repetitively move the limbs of patients. There is also a degree of variability to how it is applied, as different therapists can perform the motion in different ways. Additionally, the same therapist may perform differently from the beginning of the day to the end of the day when they are more tired.

These therapy methods can be incorporated with rhythmic auditory stimulation (RAS) to achieve additional gains. RAS consists of playing music with accentuated beats that correspond to the desired cadence of walking [4]. While the exact mechanism is still being studied, RAS has been shown to improve gait symmetry, in terms of stride time, stride length, weight bearing, and muscle activation patterns; in addition to reducing variability in stride time and length. [5–7]. However, this is not necessarily generalizable to all types of lesions or levels of impairments.

Task specific therapies lend themselves well to the incorporation of new technologies and will be further discussed in the next section.

1.2 Robot Assisted Gait Training

Robot assisted gait training (RAGT) is gaining popularity due to its ability to reduce manual effort for physical therapists, record quantitative measures of improvement, and provide patients with consistent and repeatable therapy [11]. Rehabilitation robots can reduce strain on physical therapists by reducing the

(a) Hal [8] (b) ReWalk [9] (c) Ekso GT [10]

Figure 1.1: A selection of mobile exoskeletons for gait.

repetitive manual labor, but it also requires additional training to be able to incorporate robotic systems into the physical therapists' practice. While the benefits are helpful to care providers, the ultimate value of this technology to the patients, is still under debate, as there have been mixed results in terms of outcomes [12–20]. It should be noted that for most studies that showed RAGT to have greater improvements over conventional therapy, RAGT was done in addition to conventional therapy, and/or used subacute populations. While this result may seem disappointing, it challenges our community to provide solutions to improve outcomes in the future, and this, in part, motivates the work presented in this paper.

Exoskeletons can be divided into two categories, supported or unsupported. An unsupported exoskeleton is attached to the user but has no other external grounding (not supported by a structure). This provides an individual with a near limitless workspace, but has power limitations because energy must be stored in the device or the device must be tethered with a power cord, reducing some of the advantages of an unsupported device. These also face weight limitations because the user has to support the device. This can be somewhat mitigated in gait exoskeletons because the load can be transfered to the walking surface through the stance leg. These are not externally supported, and as a result tend to have more freedom allowed at the pelvis, but this is dependent on the design. When focused on health care, these devices tend to be more assistive in nature, as these can be worn outside of the clinical setting and can aid the wearer with activities of

(a) Lokomat [30] (b) LOPES [31] (c) ALEX I [32]

Figure 1.2: A selection of laboratory based exoskeletons for gait.

daily living. HAL[1] [21, 22], Ekso[2], and ReWalk[3] are examples of unsupported exoskeletons. A supported exoskeleton is attached to the user and to a stationary reference like a wall or the floor. This is typically done to prevent the wearer from needing to support the weight of the device, but has the drawback of a limited workspace as the exoskeleton can only move a limited distance from the supporting interface. Lokomat [13, 23], LOPES [24–26], and ALEX [27–29] are all rehabilitation exoskeletons, which typically fall into this category as they can provide a more structured environment and do not face some of the challenges of unsupported devices.

1.3 Controller Strategies

Early therapy methods directed specifically at physical therapy relied on position based controllers which moved the person through a predefined path [33–36]. Devices for motor control research have found some limitations to this approach to training movements. Regardless of whether a motion was externally or self initiated, the same motor areas of the brain were activated, although self initiated movement affected a larger area of that region and produced greater inter-cortical excitability [37]. A similar study showed that training of a thumb movement was only encoded when a movement was actively rather than passively trained, as

[1] Cyberdyne Inc., Tsukuba, Ibaraki, Japan
[2] Ekso Bionics, Richmond, CA, USA
[3] Argo Medical Technologies, Ltd., Yokneam Ilit, Israel

evidenced by a change in the motor evoked potential from trans-cranial magnetic stimulation, which was only observed in the actively trained group [38]. Robotic therapy through passive movement provides the force required to complete movements, which may be necessary for severely impaired individuals, because without this level of help they may not be able to achieve the motion. However, this limits the amount of active participation by the subject as well as reduces the variability in the movement.

To address this problem new controllers were developed that would assist only as needed. These can vary from requiring the subject to initiate movement before the robot starts to apply force [39, 40], to those that provide limited guidance when following a path [41–44]. These controllers allow some level of active participation, which improves encoding of trained movements [38]. They also allow greater variability, which aids in creating generalizable motions that can be adapted to different situations [45,46]. Interestingly, high levels of variability can interfere with learning a task, so it would seem there is an optimal level which is not at the extrema [47]. Cai et al. were able to show in an animal model, that these strategies are more effective for gait training than position based controllers [48]. However, with these controllers the subject may only put in as much effort as required to complete the motion, referred to as “slacking”, and reduce their effort to increase the level of assistance from the device. In order to prevent this, controllers that adapt to “slacking” have been proposed that reduce the aid from the device if the person is relying too heavily on it [41]. However, this strategy has been primarily tested in upper limb therapy [49–51].

An alternative to the anti-slacking controllers mentioned above are error augmenting controllers. Instead of helping the user to follow the prescribed motion, these increase the amount of error, forcing the user to actively resist the device to achieve the goal. In this way, the user cannot rely on the device to aid them in completing the task and instead the user takes a more active role to reach the goal. It has been observed that for short training, protocols with forces acting opposite to the desired direction of motion can improve the outcomes [52,53]. Kao et al. also showed that error augmenting increased the rate of adaptation and reduced aftereffects during catch trials as compared to assist-as-needed (AAN) strategies [54]. This is supported to some degree by other studies showing that non-assistive forces can improve outcomes for timing error of wrist motions in healthy individuals [52], and for self selected speed and Berg Balance scores in individuals with chronic spinal cord injury (SCI) [55], although subjects in this study performed both assistive and resistive RAGT.

While there is a great deal of diversity in possible control strategies there has been limited testing to evaluate the benefits and drawbacks of each [48, 52, 54, 55]. As the field moves forward these control strategies should be evaluated against each other in impaired populations to understand the severity and type of impairment that each strategy is best suited to. Once understood, each control strategy can be applied to individuals to maximize the benefits of their treatment.

1.4 Outline of the Thesis

The next chapter will describe the ALEX II device, including the passively gravity balanced support system, the leg, actuation, sensors, and controller. The third chapter contains an experiment which addresses how the amount of freedom allowed at the pelvis affects the adaptation to new gait patterns in healthy individuals. The fourth chapter presents the design of ALEX III, a bilateral exoskeleton with 12 active degrees-of-freedom. The chapter includes a description of the supporting platform, the legs, actuation, sensors, and the controller. The fifth chapter contains the design of a novel haptic feedback strategy for gait training based on a variable damping field. In that chapter several configurations of this haptic feedback are compared to determine an appropriate configuration for future experiments, in addition to validating the controller's ability to induce gait adaptations in healthy individuals. The sixth chapter compares this new haptic strategy to two prior strategies to compare the benefits and drawbacks of each. The seventh chapter presents an overview of the design objectives for ALEX II and ALEX III, in addition to several design modifications that were made to improve performance. The final chapter summarizes the contributions of the presented work and suggests open areas in this field which build upon the findings presented in this work.

Chapter 2

Design of ALEX II: an Exoskeleton for Unilateral Gait Training

The ALEX II device, shown in Figure 2.1, was designed to replace the original ALEX I device. ALEX I could only be used on the right leg and used linear actuators which limited the range-of-motion of the leg. It also did not allow for anterior/posterior motion of the pelvis. Like the previous exoskeleton [42], ALEX II is able to apply torques at the hip and knee joints of the users, which include healthy young adults, healthy elderly individuals, and stroke survivors. The ALEX II device is supported from the rear, and attaches to the users waist, thigh, and shank. The unilateral leg can be used on either side of the subject as necessary, an improvement over the previous design. Additionally, several adjustments within the exoskeleton's leg can also be made, improving the fit to the user.

The bulk of the device is made from 6061 Aluminum, utilizing ABEC-1 bearings in all passive revolute joints. The back support is gravity balanced using springs which were custom cut from stock lengths.

2.1 The Gravity Balanced Support System

The support system allows for four uncoupled degrees-of-freedom; see Figure 2.2. Two parallel linkages provide the anterior-posterior motion (C) and superior-inferior motion (B). The two parallelogram linkages decouple these two motions and are gravity balanced. It should be noted that (B) has a noticeable amount

(a) A CAD model of the ALEX II exoskeleton.

(b) A subject walking in ALEX II.

Figure 2.1: ALEX II which provides active control of hip and knee flexion/extension. The support system is passively gravity balanced so the user does not have to bear the weight of the exoskeleton.

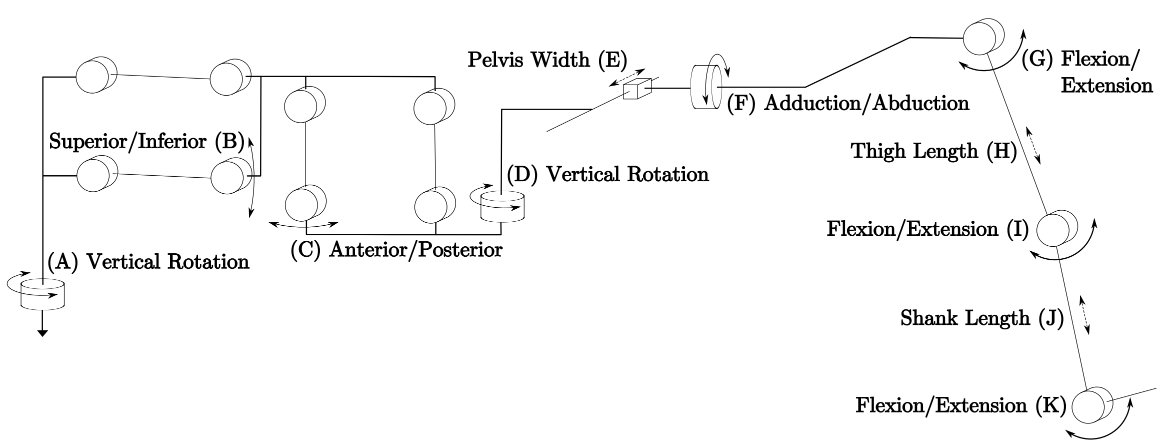


Figure 2.2: Kinematic structure of the ALEX II open chain manipulator.

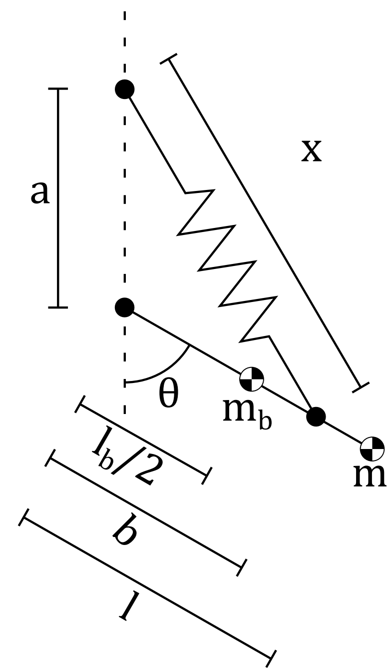


Figure 2.3: Definition of variables for Equations (2.1)

of friction, due to the tension in the cable running over the routing pulleys. Two more degrees-of-freedom provide side-to-side translation (A) and rotation about a vertical axis (D). Passive gravity balancing was accomplished using the methods previously described by Agrawal et al. [56] and will be discussed later in this section. In this device, the concept has been extended to gravity balance two degrees-of-freedom simultaneously.

The previous version (ALEX I) has only a simple parallelogram linkage similar to (B). At the end of this linkage, there are V-groove rails that provide side-to-side motion. Then, there is a revolute joint corresponding to (D). It does not have a degree-of-freedom in the anterior-posterior direction, so when its linkage moves up or down, the subject moves slightly back. ALEX I is supported by a vertical spring that is adjustable so that the subject doesn't feel the weight of the exoskeleton when standing normally. When raised above that point, the subject feels a small portion of the weight; if the subject is below that point the spring provides an upward force.

In ALEX II, neither the weight of the leg nor the back support are borne by the subject. Instead, the back support provides configuration independent gravity compensation for the device. The back support consists of a series of spring compensated links, B and C from Figure 2.2. Each of these joints has independent gravity balancing systems which work in the same way, by balancing the potential energy which comes from changes in height of the load (PE_g) with potential energy of a zero free length springs (PE_k). The model can be seen in Figure 2.3. The zero free length springs are configured so that when $x = 0$ there is no force from the spring. The load, of mass m, is at length l from the end of the supporting member. The supporting member has a mass, m_b, centered at a length of $l_b/2$ from the end of the supporting member. The spring originates from directly above the base of the supporting member at a distance of a and terminates on the supporting member at a distance of b from the base. The potential energy of the load and the spring are shown in Equation (2.1a) and Equation (2.1b), respectively. Constant potential energy is given by Equation (2.1c). The potential energy is constant and independent of configuration when $abk - mgl = 0$ and this can then be used to solve for k, Equation (2.1d). This is simplified and does not include the gravitational load of the links. However, the link's potential energy has the same θ dependence as PE_g, and can be added to Equation (2.1c) and used to solve for k, Equation (2.1e). Two parallel linkages, as seen in Figure 2.2, can be placed in series to achieve two uncoupled, gravity balanced degrees-of-freedom, since the origin of the spring on the second DOF can maintain its vertical configuration with respect to the base of the links which it is compensating.

$$PE_g = mgh = mgl(1 - cos\theta) \tag{2.1a}$$

$$PE_k = \frac{1}{2}kx^2 = \frac{1}{2}k(a^2 + b^2 - 2abcos(\pi - \theta) = \frac{1}{2}k(a^2 + b^2) + abkcos\theta \tag{2.1b}$$

$$PE_g + PE_k = const = mgl + \frac{1}{2}k(a^2 + b^2) + (abk - mgl)cos\theta \tag{2.1c}$$

$$abk = mgl \rightarrow k = \frac{mgl}{ab} \tag{2.1d}$$

$$PE_b = m_b g\frac{l_b}{2}(1 - cos\theta) \rightarrow k = \frac{mgl + m_b g\frac{l_b}{2}}{ab} \tag{2.1e}$$

2.2 The Leg Design and Functionality

In order to accommodate users with either a left or right paretic leg, the exoskeleton leg can be easily switched between the two sides. This is achieved with an easy release latch on a revolute joint in the transverse plane (which rotates when the clip is removed during side switching) and an easy release carabiner that connects a retaining spring to the adduction/abduction joint. This is an improvement over ALEX I, which could only be used on the right leg, severely limiting the subject pool which could be drawn on for study.

Similar to its predecessor, the length of each leg segment is adjustable to accommodate different sized subjects, thigh length (H) and shank length (J) are shown in Figure 2.2. The pelvic width (E) is also adjustable using a screw driven prismatic joint. Additionally, the distance to each leg segment from the exoskeleton is adjustable, not shown. This is provided because subject thigh and shank widths vary. Failure to accommodate for this difference can result in inappropriately aligned joints. There are also several different sized cuffs to allow for different leg circumferences.

The support for the hip motor allows for adduction and abduction through a revolute joint behind the subject's femoral head (F). In addition, the hip (G) and knee (I) can flex and extend in the sagittal plane, like the previous design. In the first iteration of ALEX II, there is an additional degree-of-freedom allowing medial and lateral rotation of the hip through a parallel linkage system (not shown in Figure 2.2), however due to the difficulty of holding a near cylindrical leg fixed in a near cylindrical cuff, this degree-of-freedom did not function effectively and was removed from the current version of ALEX II.

There is an encoder on the ankle which is used to measure its joint angle (K). This is a completely passive component made of lightweight polypropylene. It only measures the ankle angle and does not apply joint torque. ALEX I has a similar joint, which uses a shoe insert that requires a physical therapist to put the subject's shoe on while the subject stands in the device. Since this is rather difficult to do, the current ankle joint can be put on the subject before getting into the exoskeleton. In the current version, due to limitations in the number of available encoder inputs, the ankle encoder can be substituted for an ab/adduction encoder at the hip, depending on which joint is of more interest to the researcher.

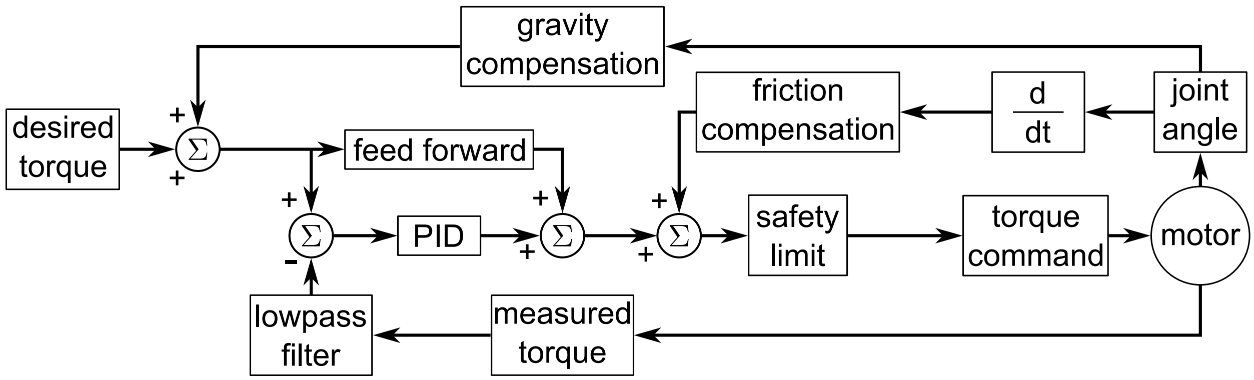


Figure 2.4: The low-level controller used in ALEX II. Note this is a torque regulator, where the set point changes according to training or free walking.

2.3 Actuation

In ALEX I, linear actuators were used. These actuators have a limited range of motion, which consequently limits the range of motion of the leg. Actuator control was also dependent on the segment length. If a subject's segment lengths were either too long or too short, the subjects could not participate in ALEX I studies, because the device was unable to produce the full range of motion characterized by normal gait. ALEX II instead uses two Kollmorgen AKM22C[1] rotary motors, geared by Thomson Micron[2] 1:50 gearboxes. These motors provide the torque commanded by the controller. Transducer Techniques TRS-1k load cells provide the feedback for closed loop control of hip and knee torques. Additionally, two ATI[3] Mini-45 six degree-of-freedom force-torque (F_x, F_y, F_z, τ_x, τ_y, τ_z) sensors mate the exoskeleton to each orthotic. These can be used for control of the robot and have the advantage of sensing the interaction force directly; thus, the inertia of the robotic leg becomes an error term and can be removed without the need to model it [29]. The motors provide a workspace to the anthropometric exoskeleton legthat is similar to a human leg, with software and mechanical stops at the limits of the leg's range-of-motion.

2.4 Control

A dSPACE 1103[4] control system is used to implement the real time control and for all data collection. The low level impedance controller has been designed to compensate for drive-train friction and gravity, as shown in Figure 2.4. When walking in the exoskeleton, but without the force field (explained in the next paragraph), the system regulates the torque at each joint toward zero. In this way, when the users apply a torque to the system, they effectively guide the exoskeleton to follow their movements.

$$F_N = \begin{cases} -K_N \times (sign(d))(|d| - D_0/2)^2 & \text{if } |d| \geq D_0/2 \\ 0 & \text{if } |d| < D_0/2 \end{cases} \tag{2.2}$$

During gait training a force field is applied to the leg. The torque control is determined by modeling a virtual spring, as used in [42]. A target ankle path is generated for the specific subject. Around this path a virtual tunnel is created, and the walls of this tunnel are characterized as a non linear spring, Equation (2.2). Here F_N is the force along the normal vector, K_N is the spring stiffness, d is the distance from the path, and D_0 is the tunnel width. When the ankle (end point) interacts with the wall of the tunnel, it is pushed back toward the target path. The properties of that virtual spring can be easily changed by the therapist to suit the needs of the subject. The tunnel width can also be set by the therapist and is generally widened as the subject learns the new gait pattern. A force tangent to the path can be applied to the leg if the subject is having difficulty maintaining walking speed. This tangent force is only applied when the subject is sufficiently close to the target path, Equation (2.3). Here F_t is the force along the tangent vector, K_T is the stiffness, d is the distance from the path as above, and D_T is the tunnel width. Tangent force is inversely related to the distance from the path, as the deviation becomes larger, the force provided is lessened. This encourages proper path following rather than focusing on overall speed, and requires subject engagement. The tunnel width and stiffness can be adjusted independently from the normal force, allowing for the tangent force to be tailored to the specific needs of a user.

[1] Danaher Corporation, Washington D.C., USA
[2] Danaher Corporation, Washington D.C., USA
[3] ATI Industrial Automation, Apex, NC, USA
[4] dSPACE GmbH, Paderborn, Germany

A secondary control strategy utilized by ALEX II is the error-enhancing strategy [54]. This works in much the same way as the AAN strategy, except the force pushes the ankle away from the path instead of towards it. This forces the person to actively correct errors rather than relying on the force to keep them near the path. For safety, a saturation limit is set on the force to prevent it from becoming too large for the person to overcome.

$$F_T = \begin{cases} K_T \times \left(1 - \frac{|d|}{D_T/2}\right) & \text{if } |d| \leq D_T/2 \\ 0 & \text{if } |d| > D_T/2 \end{cases} \tag{2.3}$$

2.5 Additional Sensors

Each shoe worn by the subject is instrumented with three Interlink Electronics FSR 406[5] pressure sensors, mounted at the heel, ball, and toe of the foot. These provide information about foot contact with the treadmill. This is used in offline data processing and can trigger a functional electrical stimulation (FES) unit when FES is used.

2.6 Summary

ALEX II is a unilateral robotic leg trainer with a configuration independent gravity balanced support system, providing three translations as well as rotation about the vertical axis. In addition to active hip and knee flexion/extension, it has passive hip abduction/adduction. Motor torques can be sensed at the gearbox output shaft, and human/robot interaction force can be measured at the human/robot interface. The features that have been added to ALEX II make it a very capable platform for evaluating how to best provide gait training. Due to this high degree of functionality, different features can now be systematically removed to evaluate their role in gait training. One example of this will be presented in the next chapter.

[5] Interlink Electronics, Camarillo, CA, USA

Chapter 3

Degrees-of-Freedom Study

3.1 Motivation

One of the major drawbacks of rehabilitation robots is their cost, which can often be decreased by reducing the complexity. The challenge is to achieve this without sacrificing quality of rehabilitative care. For strengthening exercises, it is fairly easy to target a specific muscle group with very few degrees-of-freedom, e.g., a machine at the gym. However, when trying to improve coordination, the ability to make and correct for errors in movements becomes important. This is particularly difficult with gait training because some patients may not be capable of providing the necessary motions to maintain balance, but they need to develop this skill to be successful in walking overground. The motivation behind this study was to experimentally determine the impact of limiting the anterior-posterior translational degree-of-freedom of the pelvis on adaptation to a new gait.

It has been shown that holding the pelvis fixed while walking affects gait kinematics on a treadmill [26,57]. It has also been shown that walking while wearing a rehabilitation exoskeleton alters gait kinematics and muscle activations [23,58]. Hidler states that one of the main limitations of Lokomat is the restriction it places on the movement of trunk and pelvis during walking [13]. A study with LOPES reports that limiting pelvic degrees-of-freedom with the exoskeleton has little effect on leg kinematics and muscle activation [26]. However, this study did not systematically address the issue of human learning and adaptation to new gait templates. In another study, gait adaptation, and then de-adaptation, was quantified using kinematics measured from the exoskeleton ALEX I, but did not address the issue of degrees-of-freedom [59]. It is important to point out that these results with healthy individuals may not translate directly to impaired individuals, as the impaired individuals have different abilities and training goals than healthy individuals. Similarly, the degrees-of-freedom that are optimal for stroke rehabilitation may not be the best for spinal cord injury, traumatic brain injury, or cerebral palsy rehabilitation.

This study is broken into two parts; the first is the initial study which was designed and performed to explore the differences between the amount of freedom allowed by three existing exoskeletons. It was hypothesized that increased pelvic freedom would provide a training environment more similar to normal walking and as a result would have better performance in terms of adaptation and retention. While the results do show a difference, the reason for the difference was inconclusive as multiple parameters change between two of the groups [60, 61]. The second part is an extension in which two additional groups were added to explore the effects of the individual parameters.

3.2 Initial Study

3.2.1 Participants

Seventeen healthy participants were enrolled in this study, five of which were female, the mean age of all the participants was 24 ($\pm$ 3.2) years old. More details can be found in Table 3.1. Participants self reported that they did not have any neurological or physical impairments that could affect gait. Six participants had performed a similar experiment several months prior. All participants provided informed consent and were aware that they could withdraw from the study at anytime. The experiment had open enrollment and subjects were assigned to a group based on when they performed the experiment. Group assignment occurred so that any three consecutive subjects were assigned to a different group.

3.2.2 Protocol

The experiment had three device configurations. The first had the degrees-of-freedom shown in Figure 2.2, except that D (rotation about the vertical axis) was locked; this configuration will be referred to as ALEX II. The second setup was the same as ALEX II except that joint C (anterior/posterior translation) was locked

Table 3.1: DOF Study - Subject Information

Subject	Gender	Age (Years)	Mass (kg)	Height (m)
ALEX I				
24	M	22	67	1.73
27	M	23	90	1.82
29	M	27	73	1.73
13	M	27	68	1.74
14	M	27	69	1.79
33	F	22	46	1.61
		24.7±2.6	68.8±14.1	1.74±.07
ALEX II				
23	F	30	61	1.67
26	M	22	79	1.73
17	F	26	43	1.61
30	F	18	54	1.57
8	M	26	91	1.78
15	M	25	70	1.78
		24.5±4.1	66.3±17.4	1.69±.09
Lokomat-Like				
25	M	25	84	1.80
28	M	20	70	1.83
12	M	29	75	1.75
31	M	24	65	1.76
32	F	22	51	1.59
		24.0±3.4	69.0±12.3	1.75±.09
	Study	**Extension**		
ALEX I Locked Abduction				
84	M	20	86	1.78
86	M	32	72	1.78
81	M	26	88	1.80
88	M	27	74	1.73
90	M	35	79	1.80
92	F	23	53	1.68
		27.2±5.6	75.4±12.7	1.76±.05
ALEX I Locked Lateral				
83	M	19	82	1.85
85	M	21	82	1.78
79	F	27	54	1.64
87	M	28	72	1.67
89	F	26	55	1.60
91	M	25	84	1.80
		24.3±3.6	71.4±13.7	1.72±.10
All Subjects		**25.0±3.9**	**70.2±13.5**	**1.73±.08**

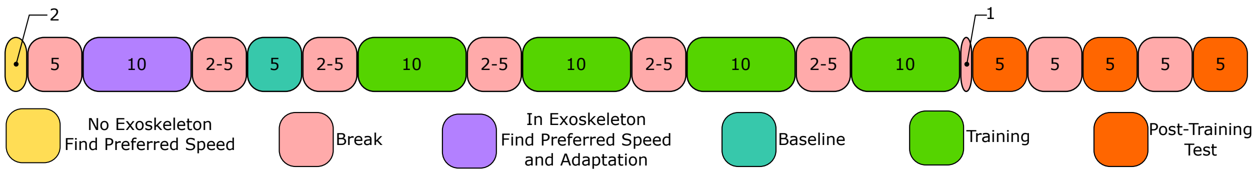


Figure 3.1: DOF Study - A graphical representation of the protocol where times are given in units of minutes. The total time was between 98 and 113 minutes.

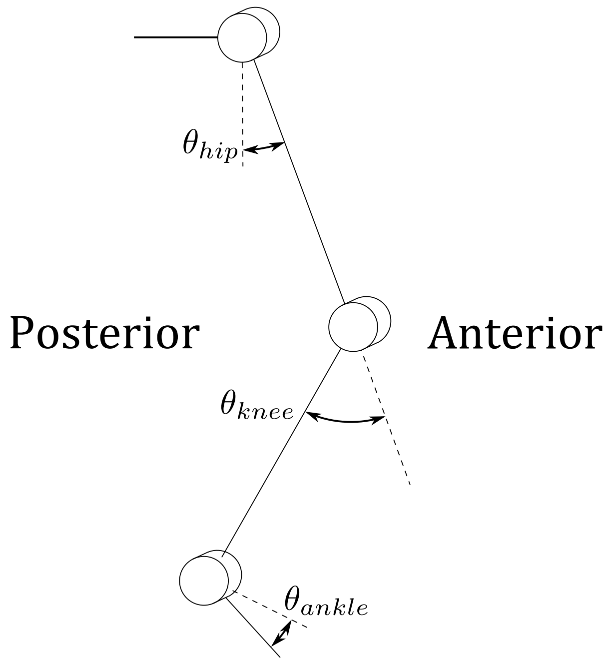


Figure 3.2: Definition of joint angles. Motion in the posterior direction corresponds to negative change in angle.

as well; this setup will be called ALEX I. It was intended to have similar freedom to the previous version of ALEX, though differences existed given that different mechanisms were used to provide the degrees-of-freedom. The final configuration was the same as ALEX I except that it had joints A (lateral translation) and F (hip ab/adduction) locked; this configuration will be called Lokomat-like. It was intended to have the same degrees-of-freedom that the Hocoma Lokomat[1] possesses, but like the ALEX I setup, it only provides similar degrees-of-freedom to the original and the motions do not correspond directly to the Lokomat. It should be noted that in all setups, joint B had a significant amount of friction due to the cable routing. Joints E, H, and J were adjusted for each participant and locked in all setups.

[1]Hocoma AG, Volketswil, Switzerland

Subjects performed the experiment in one of the three device configurations. For each configuration, the protocol in Figure 3.1 was used. The experiment consisted of a small bout of treadmill walking without the exoskeleton to find each participant's preferred walking speed; the treadmill speedometer was blinded to the participant. The device was then fitted to the subject. The participant then walked on the treadmill while wearing the exoskeleton, and their preferred walking speed was found in the same way. During this time, the motors were active and configured to only compensate for gravitational forces of the robot. Between all sessions, when the participant was given a break, they were asked to remain standing in the device. There were four ten minute training bouts. During these, the participant was asked to follow a specified foot path; we call this the target path. The target path is a modified trace of their own baseline gait. After the baseline gait was collected, each stride in the last thirty seconds was cut at its farthest anterior point and re-sampled so that all strides contained the same number of data points. The hip and knee angles are then each averaged at every data point, across all strides. Using this average, the hip and knee angles are scaled to 80% of the original value, Figure 3.3. This produces a shallower and shortened copy of the participant's original foot path, which is the target path used during training, and can be seen in Figure 3.4. The purpose of reducing the angles to 80% was to provide a gait that was significantly different than their normal gait without causing safety concerns. They were provided with an assist-as-needed force that would push their foot back toward the target as described in the previous chapter. They were also given intermittent visual feedback, as this has been found to be more effective than no visual feedback, or constant visual feedback [59]. The training period was chosen based on prior experiments. Earlier studies using 45 minutes of training found that subjects adapted their gait during this period of time. A pilot study was performed on six subjects to see if the experimental time could be reduced to the point where subjects would be willing to train in each configuration. Each subject performed all three configurations on different days after a washout period of at least two days. For each configuration only 12 minutes of training were given, and this was found to be too little time to see an adaptation. As a result, 40 minutes of training were used for the current study, as it was determined to be sufficiently long to produce adaptations and the time commitment for subjects to participate in three separate sessions for training times greater than 12 minutes would be excessive. The post training data collection consisted of three bouts without the force field. During these the subject was asked to continue to walk as they had been trained.

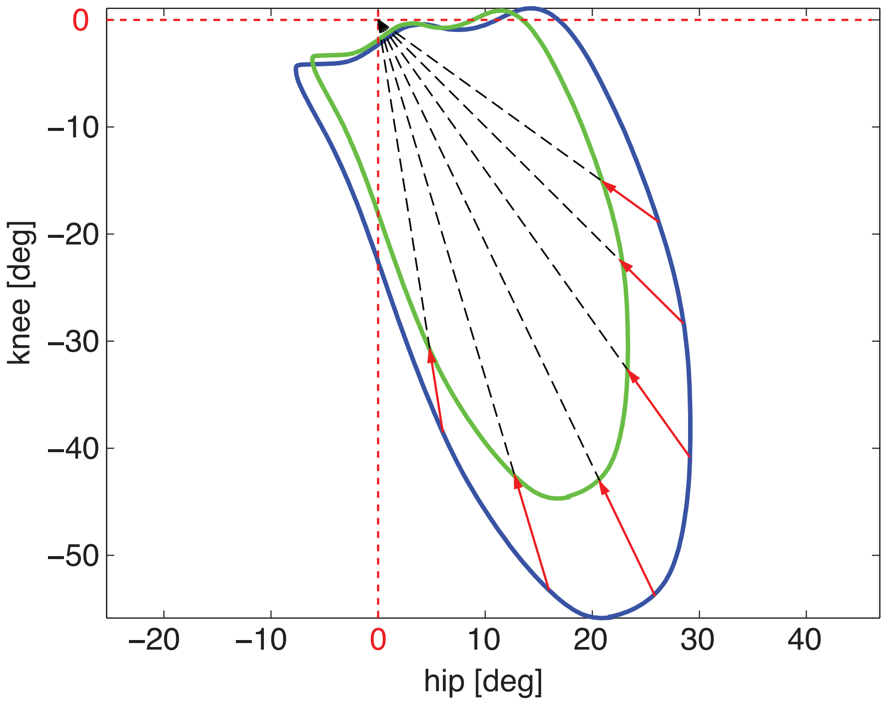


Figure 3.3: Scaling of the target path in joint space.

3.2.3 Analysis

Three values will be evaluated, normalized error area, relative normalized error area, and the reduction in comfortable walking speed. The primary measure of adaptation and retention used in this study was the "normalized error area". This area was calculated by finding the area between the actual foot path and the target template and then dividing this by the area between the original baseline template and the target template as shown in Figure 3.4. Using this measure, if a subject is closer to the target after training, the normalized error area will be less than one. Likewise, a deviation further from the target after training will result in a measure greater than one. This was calculated for every step, and then averages were made of 30 second intervals at the end of the baseline; and at the beginning, 1/3, 2/3, and end of each of the following walking bouts. To account for individual differences during the baseline data collection, the "relative normalized error area" was found by subtracting each sessions normalized error area from the baseline's normalized error area. Relative normalized error area values close to zero are near the baseline, values close to one are near the target, and negative values are further from the target than the subject was

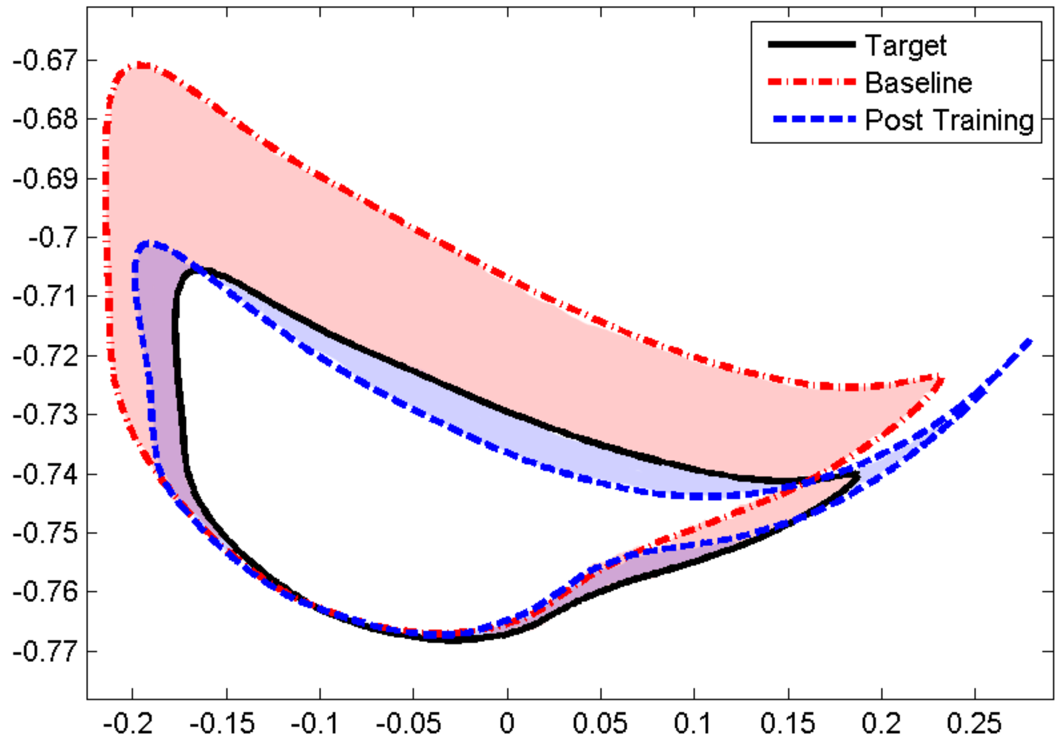


Figure 3.4: Shown are the baseline path, the target path, and the post training data in Cartesian space and represents the ankles path relative to the hip joint. The units of the axes are in meters. The shaded areas are what was used to calculate the normalized error area.
Normalized error area = $\frac{blue}{red}$

during the baseline. The final measurement was the "reduction in comfortable walking speed", which was calculated by taking their comfortable walking speed in the exoskeleton and dividing it by their comfortable free walking speed.

$$\frac{Comfortable\ Speed\ In\ Exoskelton}{Comfortable\ Free\ Speed} = Speed\ Reduction \tag{3.1}$$

For all of the analysis $\alpha = .05$. First, to confirm that all groups began with a similar baseline, the normalized error area for the baseline session was analyzed using one-way ANOVA, with configuration as the between-subjects factor.

Table 3.2: DOF Study - Normalized Error Area Post-Test

Session	ALEX I	ALEX II	Lokomat-like
Baseline	$1.05 \pm .07$	$1.06 \pm .05$	$1.03 \pm .03$
Post-Training 1	$.82 \pm .13$	$.54 \pm .24$	$.53 \pm .10$
Post-Training 2	$.87 \pm .08$	$.60 \pm .27$	$.55 \pm .16$
Post-Training 3	$.87 \pm .11$	$.54 \pm .22$	$.50 \pm .12$

Next, two-way fixed-factor ANOVA was performed on the relative normalized error area, where each subject's post-training data was pooled into one value. This was to check the role gender played in adaptation. Configuration and gender were the fixed factors. Post-hoc comparisons were performed using Tukey's HSD.

Next, to determine if the groups had an altered gait pattern following the training, each subject's data from post-training one was pooled into one value. Mixed-model ANOVA was performed between the normalized error area of the baseline and post-training one, with the session being within-subject factor and configuration as the between-subject factor. This was also repeated for the other two post-training bouts to determine if the adaptation was retained.

Finally, two-way fixed-factor ANOVA was used to examine the role of configuration and gender on the reduction in comfortable walking speed. Again, configuration and gender were the fixed-factors.

3.2.4 Results

Comparing the baselines of the groups, the null hypothesis could not be rejected, $F(2,14) = .45,\ p = .66$. From the two-way fixed-factor ANOVA on normalized error area, configuration did have a significant impact on adaptation, $F(2,11) = 8.09,\ p = .007$. However, gender did not, $F(1,11) = .07,\ p = .80$. There was also no significant interaction, $F(2,11) = .31,\ p = .74$. From the post-hoc, ALEX II and Lokomat-like configurations were significantly different from ALEX I ($p < .05$), but not from each other.

From the mix-model ANOVA, where the baseline was compared to post-training one, session did show significance, $F(1,14) = 175.58,\ p < .001$, as well as an interaction with configuration, $F(2,14) = 9.37,\ p = .003$. This was also true for post-training two and three with the main effect of session in post-training two, $F(1,14) = 93.96,\ p < .001$, and three, $F(1,14) = 153.14,\ p < .001$, and with session by configuration interaction of $F(2,14) = 6.62,\ p = .009$, and $F(2,14) = 13.05,\ p = .001$ for post-training two and three respectively. The configuration main effect did not show significance till post-training three; post-training one $F(2,14) = 3.47,\ p = .06$, two $F(2,14) = 3.44,\ p = .06$, three $F(2,14) = 6.61,\ p = .01$. Post-hoc analysis showed that ALEX II and Lokomat both performed significantly better than ALEX I, but were not significantly different from each other, and all groups showed adaptation.

Finally, for comfortable walking speed, the null hypothesis could not be rejected for configuration, $F(2,11) = 2.28,\ p = .15$, gender $F(1,11) = 3.71,\ p = .08$, or interaction $F(2,11) = .83,\ p = .46$.

3.2.5 Discussion

The initial hypothesis that more pelvic freedom would produce better performance in terms of adaptation and retention was rejected, when evaluated in the device. All groups were able to show short term adaptation and retention up to 26 minutes following the training. However, the ALEX I group did show less adaptation than the other two groups, which performed similarly. This is a somewhat unexpected result as the two with the greatest differences in degrees-of-freedom performed similarly. This suggests that the problem is not as simple as *more/less pelvic freedom is better for learning*. It appears that the combination of degrees-of-freedom needs to be explored in more detail as the results from removing degrees-of-freedom are not simply additive.

It may be that providing as much freedom as possible provides a more comfortable experience, as noted by individuals from study one who went through all configurations, and that this comfort is more conducive to training. It is also possible that by limiting the degrees-of-freedom to only those being trained, the subject is able to focus more specifically on what the robot is trying to train them to do. Because ALEX I is in between these two extremes, it may have reduced performance because the user's comfort is reduced, but it may have enough freedom to distract from the training. This is only speculation, and a greater understanding of the mechanism must be gained before the cause of this result is clear.

To discover what combination of degrees-of-freedom are optimal, we must determine the reason why the ALEX I group adapted less. Looking at muscle activation may be a good way to evaluate this, providing information on how the subjects' effort increases in the various setups, as well as how muscle synergies change as the person adapts to the exoskeleton. It may also be helpful to look at the leg motion in terms of uncontrolled manifold theory [62] to examine how a participant is trying to control the leg, as subjects may be using different control strategies based on the degrees-of-freedom allowed.

While the ALEX II group had a greater number of females than the other two groups, this does not appear to have an influence on adaptation. However, as the other two groups only had one female, the reliability of this finding may not be adequate.

While all groups did have a reduced comfortable walking speed, the null hypothesis could not be rejected for configuration or gender. This corresponds to some of the findings of Veneman et. al. [26]. This is most likely attributed to the added inertia of the robot.

One limitation of this study was the choice of which degrees-of-freedom were removed per configuration. Between the ALEX I and Lokomat-like setups, two degrees-of-freedom changed. This makes it difficult to tell if one played a more important role in the difference between those configurations or if was the combination that was important. This question will be addressed in the following section. Another limitation of this study is the evaluation process. All evaluations were done in the device with the same restrictions in degrees-of-freedom the participant was trained with. Overground or treadmill walking without the exoskeleton may be a better test, as it will show how well the training translates to real world application. In rehabilitation, the ability to perform well in the real world is much more useful to the participant than the ability to perform well in a clinic.

3.3 Study Extension

3.3.1 Participants

Twelve additional subjects were added to the study to explore the role of the intermediate DOF between ALEX I and the Lokomat-like configurations. These subjects had the same exclusion criteria and informed consent as the initial study. Their details can also be found in Table 3.1. Subjects for these two configurations were assigned to groups after the other three configurations had already been tested. Consecutively enrolled subjects were assigned to alternate groups.

3.3.2 Protocol

The protocol was the same as the initial portion of this study. The additional groups had one additional DOF locked from the ALEX I configuration and are called locked abduction, in which (F) from Figure 2.2 is locked; and locked lateral, in which (A) from Figure 2.2 which provides a mainly lateral motion is locked.

3.3.3 Results

For normalized error area, Mauchly's Test of Sphericity indicated that the assumption of sphericity had been violated, $\chi^2(5) = 69.54, p < .001$. After the Greenhouse-Geisser correction was applied, each group showed a statistically significant difference for session, $F(1.29, 31.00) = 27.25,\ p < .001$. Unfortunately with the addition of two additional groups the power of the study is reduced and statistical significance is lost for the session by configuration interaction, $F(5.17, 31.00) = 1.13,\ p = .366$. The normalized error area for each group can be seen in Figure 3.5. The locked lateral group did have a higher baseline error. If this is compensated by looking at the relative normalized error area to account for this initial offset, the trend can be made more apparent, Figure 3.7. However, the relative normalized error area did not show significance for either session, $F(1.38, 32.00) = .88,\ p = .388$, or session by configuration interaction $F(5.50, 32.00) = 2.25,\ p < .067$. Greenhouse-Geisser correction was applied as Mauchly's Test of Sphericity, $\chi^2(2) = 13.95, p = .001$, indicated the sphericity assumption had been violated.

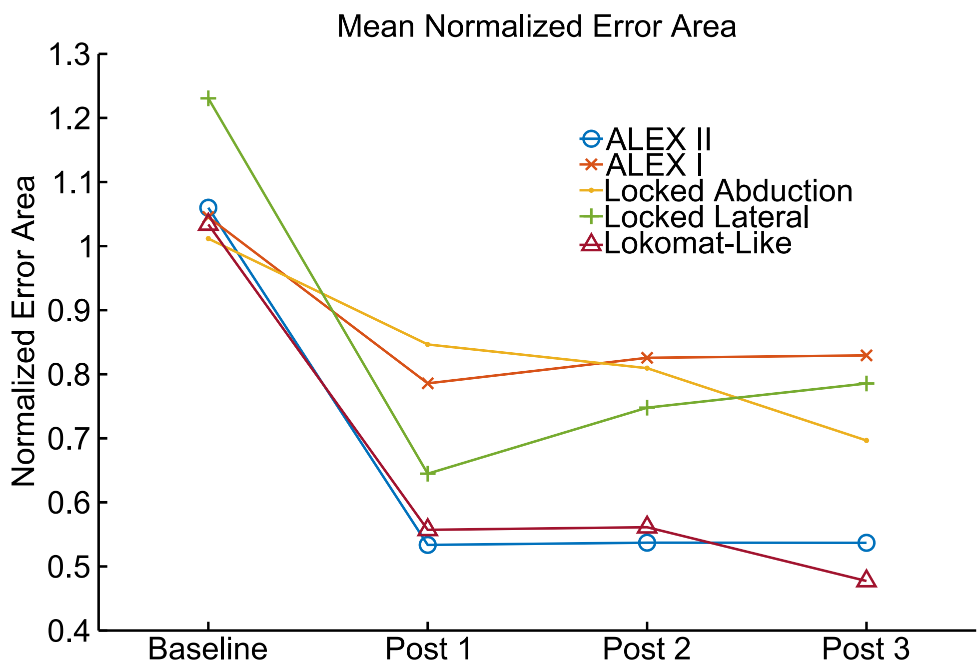


Figure 3.5: DOF Study - Line Plot of Normalized Error Area for Each Group.

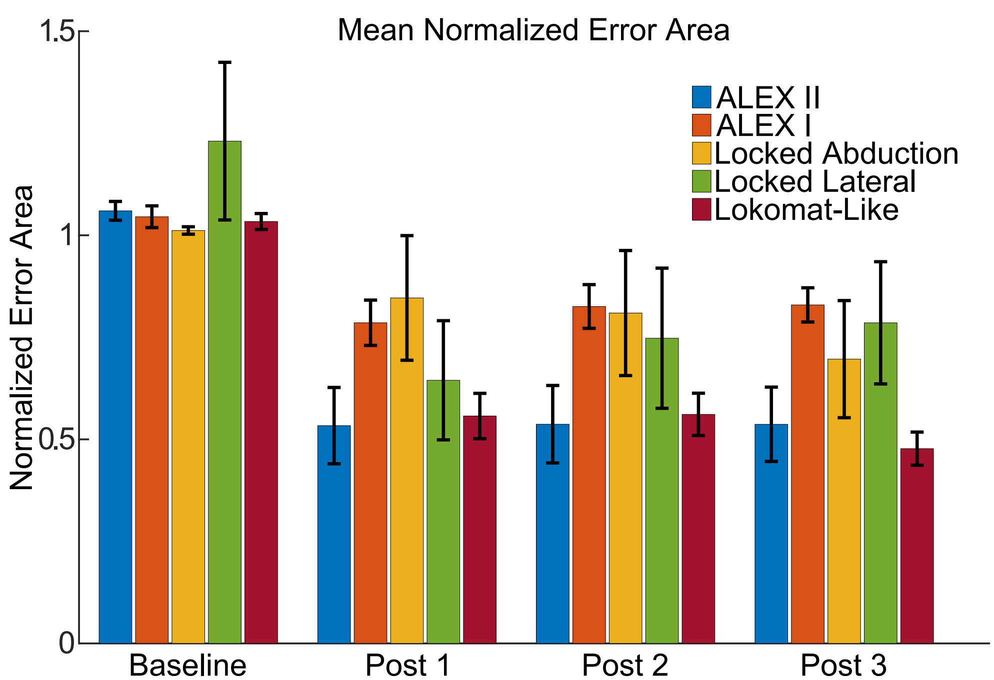


Figure 3.6: DOF Study - Bar Plot of Normalized Error Area for Each Group.

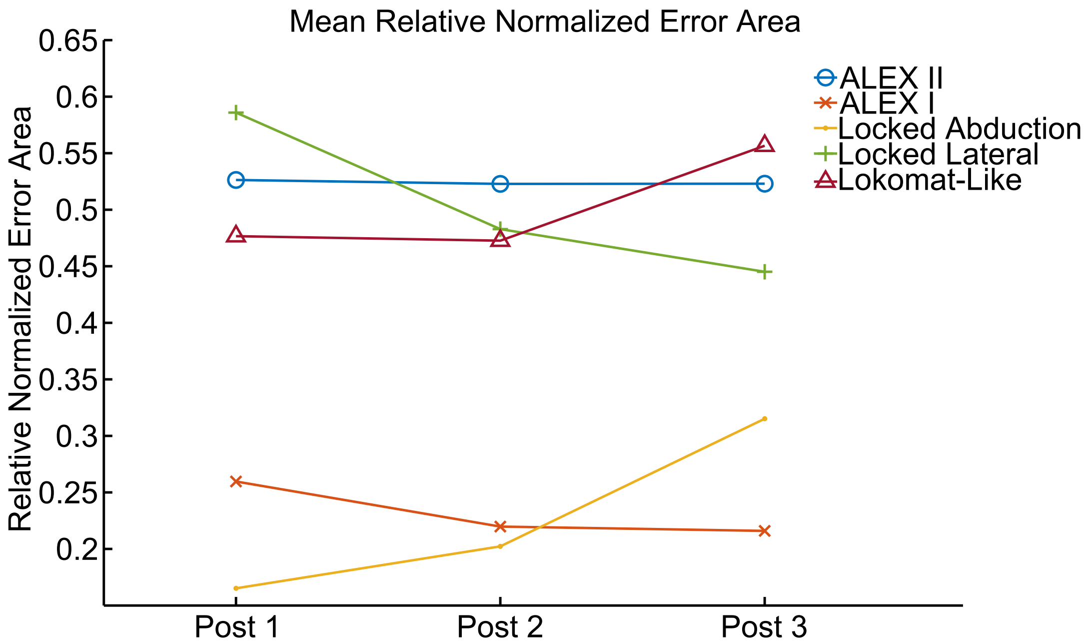


Figure 3.7: DOF Study - Line Plot of Relative Normalized Error Area for Each Group.

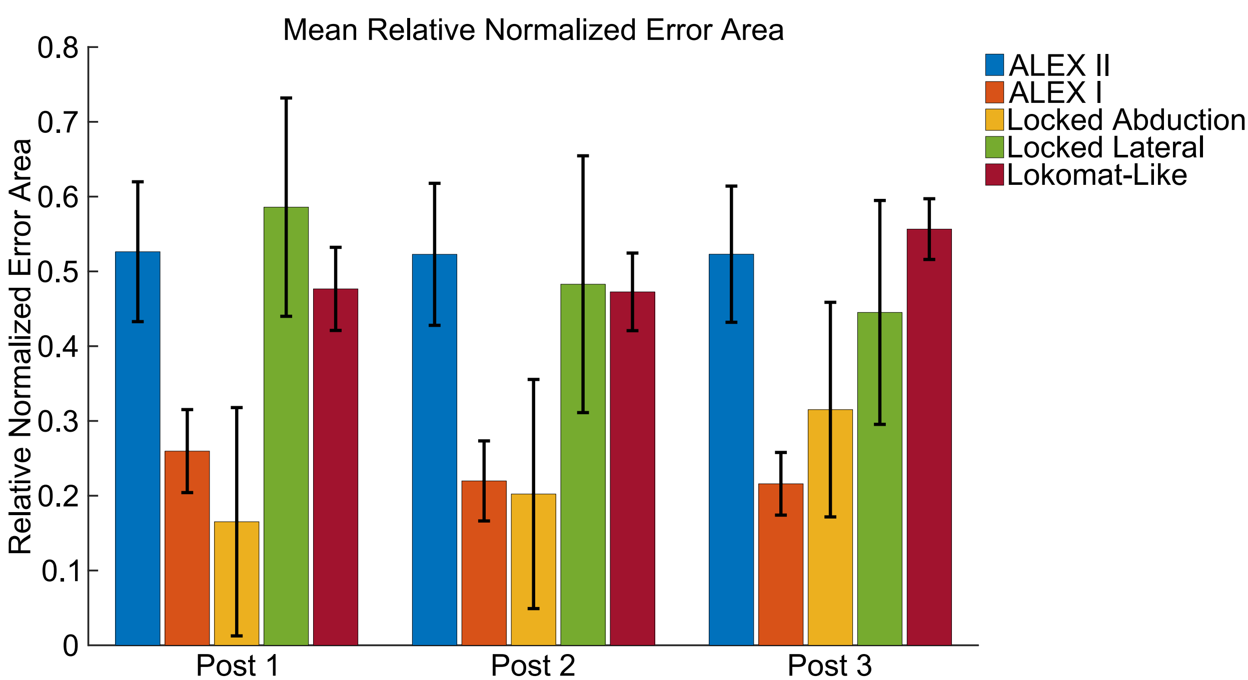


Figure 3.8: DOF Study - Bar Plot of Relative Normalized Error Area for Each Group.

3.3.4 Discussion

While the results are not statistically significant, trends can still be viewed. The results indicate that restricting the ability to ad/abduct the leg plays a lesser role in the adaptation to new gait patterns than locking a pelvic DOF. It is still unclear at this point if locking the lateral motion alone would have a similar effect to locking the anterior/posterior motion alone, but the combination of the two appears to be why the Lokomat-like configuration had the performance it did. It is hypothesized that the reason partially restricted pelvic motion limits the degree of adaptation is that it creates a more challenging balance control task, which acts as a secondary task related challenge. Alternatively, the balance task may invoke a reflex response, interfering with the training.

The extension still suffers from the limitations of the initial study, but was able to shed light on which aspect of allowed motion was the main contributor to the difference between ALEX I and the Lokomat-like group. It is worth noting that these results might differ for a bilateral device. With the unilateral device, one leg was still unrestricted, and could ad/abduct to control the lateral motion. A bilateral device where both legs no longer have the ability to ad/abduct would lose the ability to make lateral adjustments, so limiting ad/abduction in that case may be essentially the same as locking lateral motion.

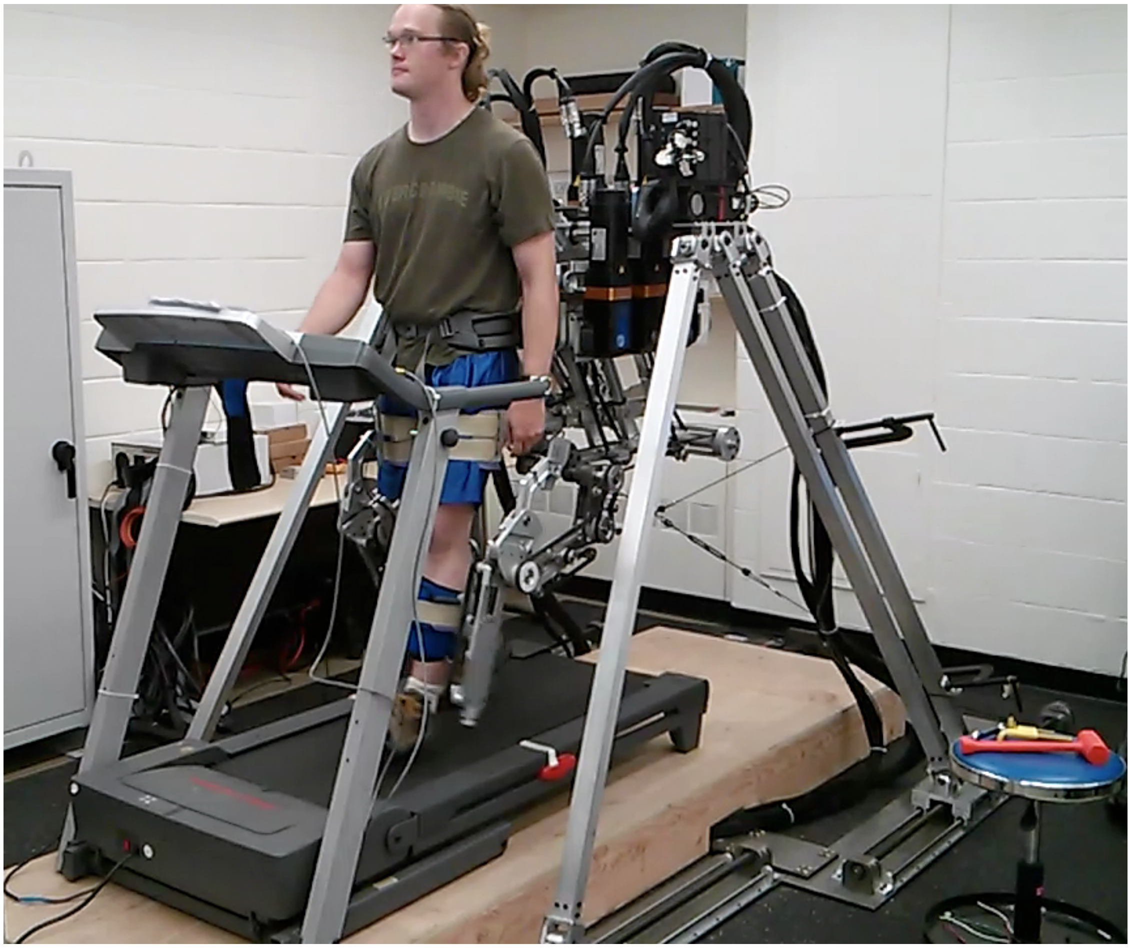
Chapter 4

Design of ALEX III: an Exoskeleton for Bilateral Gait Training

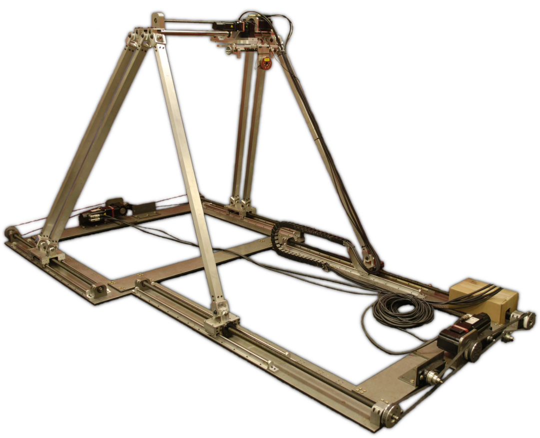
ALEX III, Figure 4.1, is a bilateral device which provides the ability to track and control both legs. Each leg is capable of abduction/adduction at the hip, and flexion/extension at the three main joints of the leg. The pelvis is capable of three translational motions as well as rotation about the vertical axis. A majority of the components of the exoskeleton are behind the user, to allow for natural arm swing, creating a more comfortable experience. Additionally, the motors have been moved from the leg to the hip, reducing the leg's moment of inertia. All degrees-of-freedom are active, creating a richer environment to test control strategies and mechanical restrictions.

4.1 Support System

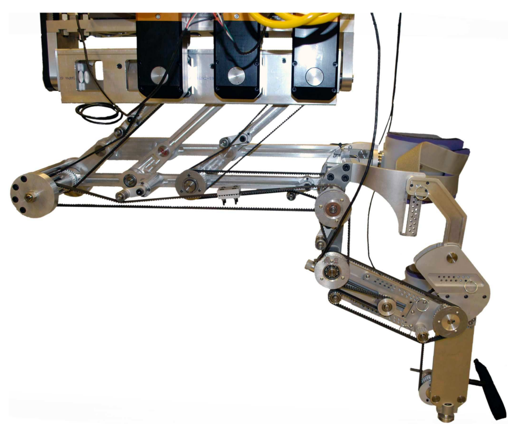
The vertical and anterior/posterior motions of the support platform are enabled by four legs. The rear legs consist of a parallel linkage, with Cardan universal joints at the top and the bottom (Fig. 4.2). Due to the nature of the parallel linkages, each rear linkage is equivalent to a *virtual leg* that runs parallel to the rear linkages and passes through the top of the front leg. The top of each leg is independently capable of reaching a spherical shell. By connecting the top of the front and virtual legs together - as is done on the left and right sides of the platform - the reachable space is restricted to the intersection of the front and virtual legs'



(a) ALEX III overall design



(b) ALEX III support system



(c) Single leg of ALEX III

Figure 4.1: ALEX III overall design and subassemblies

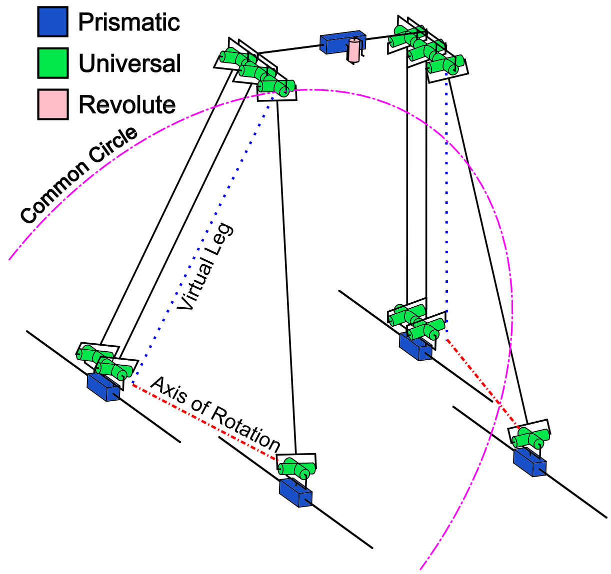


Figure 4.2: Kinematic diagram of the support system of ALEX III.

shells, i.e. a *common circle*. This common circle lies in the radical plane of the two spheres, with its center lying on the axis that connects the bases of the two legs (*axis of rotation* in Fig. 4.2), and with a radius equal to the height of the triangle formed by the same axis and the two legs.

The left and right sides of the platform are connected to each other with rails. If the radical planes of both sides are the same, then the structure would behave like a four bar linkage in that plane. However, by moving the front prismatic joints closer together, distinct radical planes are created, thereby fixing the structure for any given position of the lower prismatic joints. The system can then be modeled as a PRRRP kinematic chain composed of the projection of the spatial chain onto the sagittal plane. Moving all the lower prismatic joints forward or backwards creates anterior and posterior motion, respectively. Increasing or decreasing the distance between the front and rear bases produces inferior or superior motion, respectively. On the rails connecting the left and right side of the platform rides a prismatic joint that produces lateral motion of the pelvis. This also carries a revolute joint providing rotation about the vertical axis. The exoskeletal legs and the wearer's belt are attached after this joint.

An alternative design solution would be to use two planar PRRRP kinematic chains, as proposed in [63, 64]. However, universal joints have a significant advantage. The moment arm created by the length of the support legs is substantial: if the joints were a single revolute joint, any lateral loading at the top of the support system would result in an extremely large moment at the base of each support leg. Conversely, by using universal joints, the ends of the support legs only see moments about the longitudinal axis of the leg, so they act mainly in compression. This allows for the use of smaller components, reducing mass and moment of inertia. The support legs themselves are angled inward to help take lateral loading and increase stability.

4.2 The Legs' Design and Functionality

The leg design requires a clever mechanism to actuate the hip with two degrees-of-freedom while keeping the mechanism behind the person. Typically, hip actuation is done by having one axis of rotation behind the hip joint to provide adduction/abduction and a second axis of rotation perpendicular to that to provide flexion/extension through a motor colocated with the hip joint. This causes a problem with arm swing because the flexion/extension mechanism sits next to the person. Our goal is to prevent this. The new design, Figure 4.1c, uses a parallel linkage, similar to what provided medial/lateral rotation of the leg in ALEX II, to provide hip flexion/extension. This is done by creating a virtual joint center at the subject's hip joint. Adduction/abduction is achieved in the typical way described above.

To control the knee, a four bar linkage is used to help keep the robot behind the subject and the motors at hip level. Originally, all links were designed to be of fixed length, except the ones needed for thigh and shank length adjustment. However, this creates interference problems or singularities over the range of motion, for some segment lengths. To prevent this, a second link was made adjustable, creating another parallel linkage. This link's adjustment scale is not linear because the thigh length adjustment, which is linear, is not along the length of this new adjustable link. The robot is beside the user's leg at the knee to eliminate the weight and complexity associated with moving it behind the subject. This is low enough to allow arm swing.

The ankle joint provides plantarflexion and dorsiflexion. The attachment point to the user is the top of the shoe so the robot can push or pull the foot while allowing inversion and eversion. The thigh and shank each have cuffs that attach to their corresponding segments.

The motors are all located at the hip level of the leg so that only the adduction/abduction motor has to support their load, minimizing the overall moment of inertia. A transmission system is needed to actuate the joints because the motors are located away from the joints they control. The segment lengths vary, so the transmission needs to be flexible enough to accommodate the range of adjustments. For this reason, timing belts were chosen. Using the parallel linkage, the hip joint is controlled without the use of belts. The knee requires two belts that run along the hip parallelogram to the most proximal joint of the knee linkage; these belts run over fixed distances. The actuation for the ankle joint is more complicated. Like the knee, its belts start by running along the hip parallelogram. Then, belts must span the knee linkage, which requires an adjustable length. Finally, a belt must connect to the ankle joint across the adjustable shank length. The total number of belts needed to actuate the ankle is five.

4.3 Actuation

In total, there are twelve actuated degrees-of-freedom, four for the support system and four per leg. All of the motors used are from Kollmorgen[1]. The support system uses gearboxes and ballscrews from Thomson Linear[2]. The legs use gearboxes from Parker Hannifin[3]. The belts used are from Gates[4]. Motor and gearbox selection was done using dynamic simulation to determine motor torque and speed requirements over walking gait.

4.4 Sensors

Each motor has a built in position sensor to determine orientation. There are additional foot sensors to determine what part of the foot is in contact with the ground. Both legs have two torque transducers from Omega[5]. One is placed at the output shaft of the adduction/abduction gearbox. The other is placed between the robot ankle and the connection to the user. There are also five 6-axis force/torque sensors placed at each thigh and shank cuff, as well as at the connection between the support system and the user's pelvis. These are from ATI [6].

[1] Danaher Corporation, Washington D.C., USA
[2] Danaher Corporation, Washington D.C., USA
[3] Parker Hannifin Corporation, Cleveland, OH, USA
[4] Gates Corporation, Denver, CO, USA
[5] OMEGA Engineering, INC., Stamford, CT, USA
[6] ATI Industrial Automation, Apex, NC, USA

4.5 Control Hardware

A modular real time controller from dSPACE[7] is used to control the robot, which gives enough flexibility and power to control the system. It uses a control strategy similar to that used in ALEX II, but with improvements to the low level controller for better performance.

4.6 System Response

4.6.1 The Support System

To evaluate the platform, a single subject walked on a treadmill at 1.5 miles/hour (.67 m/s), while the support platform was worn without the exoskeleton's legs. The speed was selected from the subject's comfortable walking speed while wearing ALEX II. The motion of the end effector was recorded using the motors' encoders and the platform kinematics. The interaction forces were recorded through the force/torque sensor at the end effector.

The interaction forces were filtered using a 12 Hz Butterworth filter. The data was cut into half gait cycles when the vertical displacement reached a local maximum, roughly corresponding to mid-stance. This point was selected because no foot contact information was available, and the horizontal displacement occasionally had two local maxima per half cycle. The minimum, maximum, and root mean squared (RMS) interaction forces were found for each cycle and the mean and standard deviation over the cycles were calculated (Table 4.1). To account for the subject's changing position on the treadmill during the data collection, each cycles' position data was offset by its average value so that each cycle had a mean position of zero. The half cycles were then subsampled so that each had the same number of points. The corresponding points were averaged across the cycles.

LOPES is the only other exoskeleton with published results on interaction forces using force control. In the reference, when excited by hand at approximately 1 Hz with a 10 cm amplitude, and at approximately 3 Hz with a 2 cm amplitude the forward/backward translation had interaction force of 40 N and 100 N

[7] dSPACE GmbH, Paderborn, Germany

Table 4.1: Interaction Force in the Sagittal Plane

Vertical	Max (N)	21.23 ± 3.63
	Min (N)	-16.40 ± 7.68
	RMS (N)	12.29 ± 0.00
Horizontal	Max (N)	17.09 ± 3.86
	Min (N)	-20.30 ± 4.14
	RMS (N)	11.29 ± 0.00

respectively [25]. The vertical motion was not evaluated in LOPES, as it is passive. For ALEX II, there was no force sensor at the pelvis, as a result no interaction forces were previously recorded that can be used for comparison.

4.6.2 The Leg

As a first evaluation of the controller performance of the leg, the closed-loop disturbance rejection bandwidth was inspected by recording the force responses to externally imposed movements [65]. In this experiment, the leg was controlled in *zero-interaction mode*, and each joint was operated separately while all the others were locked. Friction compensation was not active although gravity compensation was. Sine-like movements were manually provided at one of the three robot/human interfaces: the ankle lever (ankle joint tests), the shank cuff (knee joint tests), and the thigh cuff (hip flexion/extension and hip adduction/abduction). Two sets of experiments were conducted. In the first one, the ROM of the movements and their frequency ($f \approx$ 1Hz) resembled those expected in normal walking. In the second set, the stability of the controller was checked by imposing smaller movements at a higher frequency ($f \approx$ 3Hz). Each experiment (i.e., joint/frequency combination) lasted approximately 60s, although a reduced time interval was selected for data analysis (t_{1Hz} = 30s, t_{3Hz} = 20s). Fig. 4.3 shows the imposed movements and the corresponding torque responses for the high frequency tests.

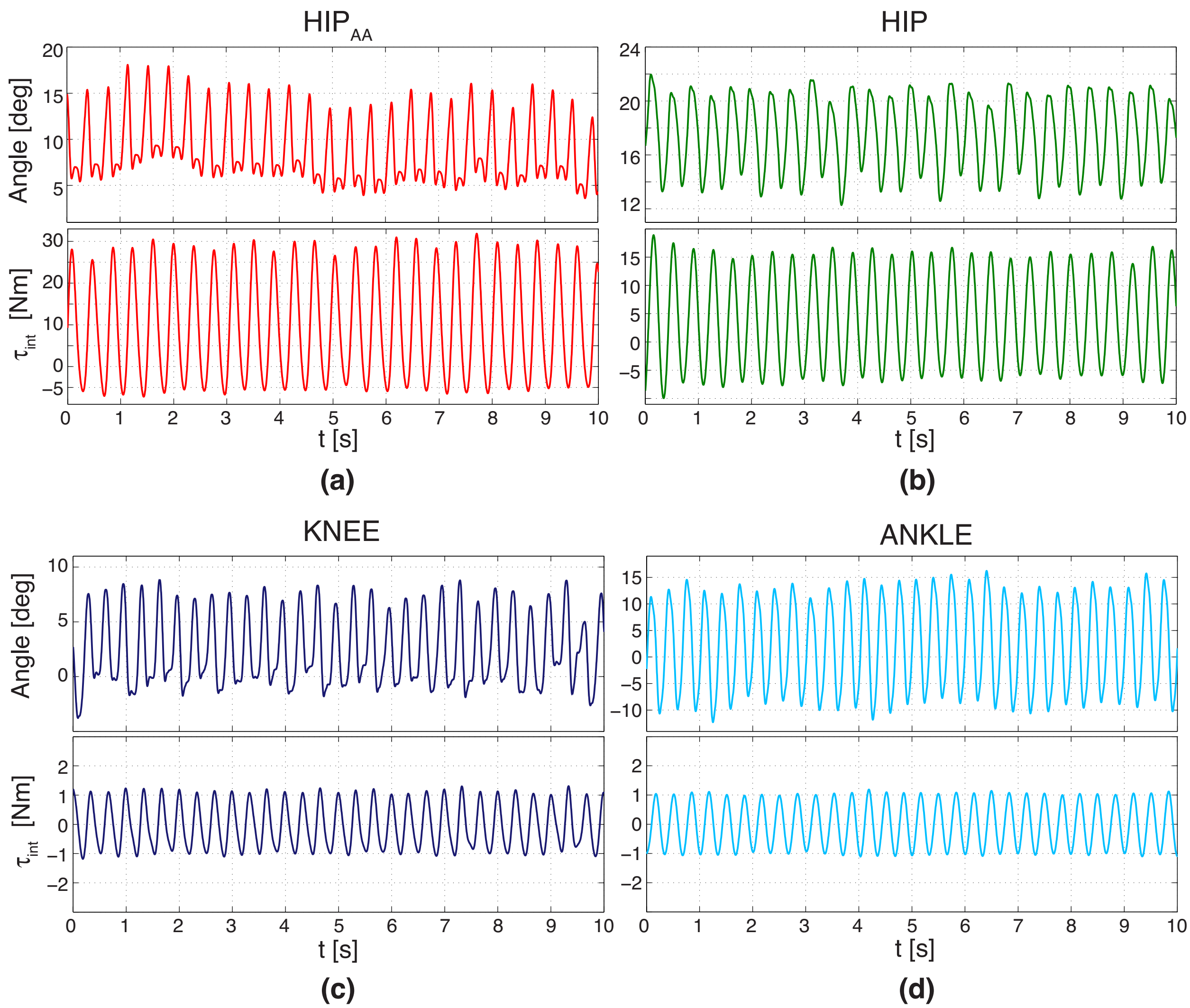


Figure 4.3: Interaction torques measured while imposing a sine-like motion to each joint separately ($f \approx 3$Hz): hip adduction/abduction (a), hip flexion/extension (b), knee flexion/extension (c) and ankle plantar/dorsiflexion (d).

Table 4.2: ALEX III Leg Response Results

	HIP A/A	HIP	KNEE	ANKLE
$ROM_{1Hz}[°]$	30	50	70	50
$\tau_{max,\ 1Hz}$[Nm]	5.31(0.63)	0.79(0.12)	1.31(0.16)	1.32(0.10)
$\phi_{0,\ 1Hz}[°]$	52(5)	45(3)	5(3)	111(3)
$ROM_{3Hz}[°]$	10	7	10	20
$\tau_{max,\ 3Hz}$[Nm]	28.42(2.17)	15.03(1.03)	1.31(0.14)	1.16(0.05)
$\phi_{0,\ 3Hz}[°]$	154(14)	135(9)	174(9)	334(10)

Table 4.2 shows the averaged peak torques and the averaged phases between position and force peaks, plus their standard deviations. The interaction was measured by converting the wrenches recorded at the active sensor to equivalent torques at the motor joints. The difference in the response of the different joints can be explained as a function of the actuation method and the amount of inertia the joint must move. For the knee and ankle joints, the controller manages to keep the interaction torques small (i.e., below 1.35Nm) for both the fast and normal-speed motions. This results from the low inertia of the belt system used in these joints. In contrast, the hip flexion/extension joint results show very small interaction torques at 1Hz, but show a larger interaction torque at 3Hz, though it is still within a suitable range. Presumably, this is due to the dynamic loads resulting from the rigid parts of the structure moving during hip flexion/extension, which are less of an issue with the belt driven joints. Similarly, the hip adduction/abduction motor must compensate for the dynamic loads of the entire leg structure (comprising the three other motors). For this reason, peak interaction torques reach almost 29Nm in high-speed movements, which would be exacerbated by placing a larger mass distally. This validates the choice of moving the motors off the distal portion of the leg. In terms of phase response, at the knee and ankle, interposing flexible transmissions between the motor shaft and the corresponding driven link increases the phase shift. This may be noticed by observing that the lag is smaller for the hip motor - which is connected to the thigh cuff through rigid members only - than it is for the other joints.

4.7 Summary

ALEX III is a bilateral robotic leg trainer with 12 active degrees-of-freedom. This provides control of the support system's three translations and rotation about the vertical axis, as well as control of flexion/extension of the three main joints of the leg and hip abduction/adduction. The ability to control each of these joints allows for the ability to test previously unexplored topics, such as the affect of pelvic perturbations on gait training, or how ankle forces can play a role in gait training. The next chapters will present the use of this device to evaluate a new type of haptic feedback and compare it to previously used methods.

Chapter 5

Damping Controller Design and Validation

5.1 Motivation

The new control paradigm utilizes a viscous damping field. The uniqueness of this new control is that it does not provide direct directional feedback, and utilizes subjects' tendency to minimize their effort in order to guide them to the target path. As a result, the user cannot rely on an external force to correct their movements, as with assistive controllers. Also, subjects do not have to actively push towards the path against the force to reduce the error, as in the error-enhancing controller. The viscous damping field, instead, provides a resistance in the direction of motion based on the error from the desired path and the ankle velocity.

From a motor learning perspective, the viscous damping field represents a more descriptive form of knowledge of performance by describing the magnitude of the error but not providing any information on how to correct it. Conversely, error enhancing and reducing strategies are consistent with prescriptive knowledge of performance by providing information as to what needs to be done to correct the error through forces related to the direction of error [66, 67]. It is debated which type of feedback is best, but is generally agreed that it depends upon the task and skill level of the individual [68–70]. Novices tend to benefit more from prescriptive feedback as they may not have enough experience to come to a safe or optimal movement strategy, and instead learn by being told what should be done to improve performance. More experienced individuals who are allowed to use discovery through exploration have been shown to have faster reaction times, and better accuracy when transfer tests occur in environments with stressors, such as during a competitive game [68, 69]. This is thought to be due to a lower cognitive load from more intrinsic learning. This improvement is because they do not need to recall specific information as to how to complete the task. For gait training, this may create better transfer to the real world as walking occurs in cluttered and dynamic environments which can create more stress than a controlled laboratory environment.

For the upper limb, some work has been done in the area of damping for movement training. A device with two DOFs has been developed for haptic feedback at Osaka University. It is designed to be used with the hand and uses electrorheological clutches to transmit force from the actuators. It can simulate virtual objects with spring-damper properties and has been tested with path following tasks where the spring force is determined by the position error and the damping force is constant [71]. It has also been used to simulate the interface with a virtual wall, but forces were again position based despite being realized through an electrorheological clutch [72]. The MIT-Manus has been used with viscous force fields for perturbation, however, these did not behave as dampers as the force applied was not in the direction of travel but rather at a predefined angle [73–75].

For gait, Wu et al. have developed a cable driven device capable of resisting or assisting in gait. This uses the position error to create an assistive force and the velocity error to create a resistive force. However, the assistive and resistive coefficients are determined by subject tolerance and are not variable with error. The resistive force is also based on velocity error and not the foot's actual velocity, meaning it is minimized when the foot is traveling at the desired velocity rather than along a desired path [55, 76]. The Lokomat has also been tested with resistive damping based on joint velocity. The damping coefficient was fixed [77] or based on the maximum voluntary contraction and walking speed [78], and produced a constant damping coefficient throughout each session. To the best of the authors' knowledge, the use of variable damping controllers based on position error is largely unexplored.

5.2 Design

The current controller is built from prior work using force tunnels [42, 54]. It has now been modified so that as the distance outside the tunnel increases, the damping coefficient applied to the foot increases till it reaches saturation, Equations (5.1), (5.2), (5.3). Here $\mathbf{F}$ is the Cartesian force vector applied at the ankle

Table 5.1: Damping Parameter Study - Subjects Information

Group	Sex	Age (yrs)	Mass (kg)	Height (m)
Lin Low	6 M, 2 F	24.1±5.1	73.0±9.9	1.80±0.06
Lin High	5 M, 3 F	27.6±4.7	81.5±17.8	1.79±0.08
Par Low	6 M, 2 F	25.5±4.4	80.0±21.1	1.77±0.09
Par High	5 M, 3 F	21.6±2.6	69.1±9.1	1.75±0.08

and $\mathbf{V_{ankle}}$ is the Cartesian velocity vector of the ankle. D_0 is the tunnel width and d is the distance of the current ankle point from the path, and B_1 and B_2 are the damping coefficient gain for the linear and parabolic fields respectively. B is the damping coefficient and is calculated from Equations (5.1), or (5.2) depending on the group, and B_{max} is the saturation point for the damping coefficient. The saturation point was set, for safety, to a value at which healthy individuals would still be able to complete a step. The high and low values for B_1 and B_2 were selected so that saturation was reached the same distance from the tunnel wall as the high and low stiffnesses used previously in the AAN training paradigm 5.1. Therefore, the saturation point of B_1 and B_2 for the high setting occur at the same location as the saturation point of the high stiffness setting of the AAN and error-enhancing controller. This also follows for the low setting.

$$B_{linear} = \begin{cases} B_1 \times (|d| - D_0/2) & \text{if } |d| \geq D_0/2 \\ 0 & \text{if } |d| < D_0/2 \end{cases} \tag{5.1}$$

$$B_{parabolic} = \begin{cases} B_2 \times (|d| - D_0/2)^2 & \text{if } |d| \geq D_0/2 \\ 0 & \text{if } |d| < D_0/2 \end{cases} \tag{5.2}$$

$$\mathbf{F} = \begin{cases} -B \times \mathbf{V_{ankle}} & \text{if } B \leq B_{max} \\ -B_{max} \times \mathbf{V_{ankle}} & otherwise \end{cases} \tag{5.3}$$

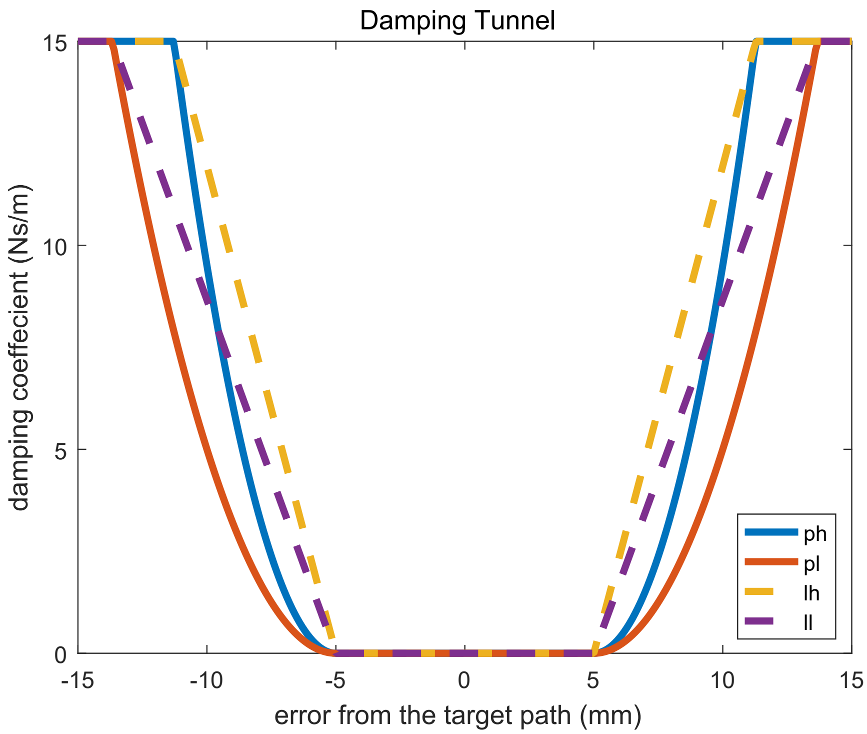


Figure 5.1: Damping fields used in the damping parameter experiment.

5.3 Protocol

Thirty-two right leg dominant, healthy individuals, with no neurological or physical impairments that would affect their ability to walk in the device or adapt to new gait patterns were recruited, Table 5.1. Each subject performed a single testing session using the ALEX III device, Figure 5.2 [44, 65]. Subjects were separated by gender, and randomly assigned to one of four groups. Each group had either a linear or parabolic shaped field, and either high or low strength. This was done to determine which setting should be used for future experiments and to evaluate the effectiveness of the controller. It was hypothesized that a more gradual change in the damping field would make it easier for subjects to feel the damping gradient and use it as a guide in following the footpath, and would additionally allow them to feel the gradient at a larger distance from the path prior to saturation.

Each session, Table 5.2, began by finding the subject's comfortable walking speed in the device. Next, the subject walked for ten minutes to acclimate to walking in the device. After a break, a five minute baseline bout was performed to record their normal walking in the device. The average baseline footpath

Figure 5.2: Researcher walking in ALEX III.

was then modified to create a footpath with a 50% increase in step height in early swing. This new path was the target path used in training. Subjects then performed four ten minute training bouts with the damping field described above applied to their left leg and intermittent visual feedback, alternating on/off every 2.5 minutes (50% frequency). Each training bout was followed by a thirty second catch-trial, a one minute break, and one minute mid-test without force or visual feedback. The last catch trial was followed by a one minute break and a five minute post-test instead of a one minute mid-test. Two more post-tests followed with five minute breaks separating them. These post-tests were performed in the device, as a pilot study (N=8) indicated that the training did not transfer to overground walking. During training, subjects were given verbal encouragement when they were performing the task well. Subjects were asked to walk in the way that they were trained for catch-trials and mid-tests. Subjects were given no specific directions how to walk during post-tests.

Table 5.2: Damping Parameter Study - Protocol

Session	Time
Adaptation	10 min
Break	2-5 min
Baseline	5 min
Break	2-5 min
Training 1	10 min
Catch 1	30 sec
Break	1 min
Mid-test 1	1 min
Break	2-5 min
Training 2	10 min
Catch 2	30 sec
Break	1 min
Mid-test 2	1 min
Break	10 min
Training 3	10 min
Catch 3	30 sec
Break	1 min
Mid-test 3	1 min
Break	2-5 min
Training 4	10 min
Catch 4	30 sec
Break	1 min
Post-test 1	5 min
Break	5 min
Post-test 2	5 min
Break	5 min
Post-test 3	5 min

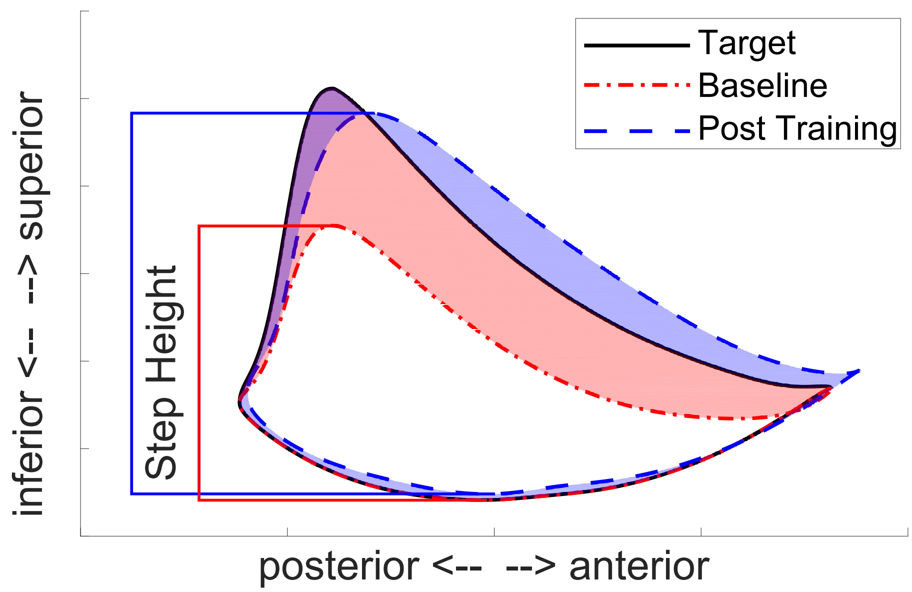


Figure 5.3: Areas and heights used for calculations of the Normalized Error Area and Normalized Step Height. $NEA = BlueArea/RedArea$, $NSH = BlueHeight/RedHeight$

5.4 Data Analysis

Analysis was performed on both the normalized error area (NEA) and the normalized step height (NSH) of the mean path for the session, Figure 5.3. The normalized error area is the area between the average path for the session and the target path, divided by the area between the baseline path and the target path. This provides a measure of how well the overall path was followed. Values closer to zero indicate that the target path was followed more closely, and values less than one indicate that the subject was closer to the target path during post-training than during their baseline session. The normalized step height takes the maximum ankle height minus the minimum ankle height of the session's mean path during post-training and divides it by the same value calculated from the baseline path. This gives a measure of how well the subjects were able to achieve the dominant feature of the path, i.e. the increase in step height. A value of 1.0 indicates that the step height remained the same as baseline, whereas a value of 1.5 indicates that the step height increased to become the same height as the target path. One-sample t-tests were performed on both dependent variables by pooling the values for all three post-tests for each subject to evaluate if their mean was significantly different from 1.0, and the Bonferroni-Holm correction was applied. One was used for both tests as it indicates baseline performance. Next, repeated measures ANOVA was performed on the post-test sessions with session as the within-subject factor, and strength and shape as the between-subject factors. If Mauchly's Test of Sphericity indicated that the sphericity assumption had been violated, the appropriate correction was applied. This test was to indicate if there was any degradation of performance over time, and to determine if there was an effect of strength or shape. For all tests, $\alpha = 0.05$.

5.5 Results

The results of the one sample t-test can be found in Table 5.3. The groups with high damping coefficient gains showed statistically significant differences from 1 in normalized error area. All groups showed statistically significant differences from 1 in normalized step height. The normalized error area and normalized step height can be seen in Figure 5.4 and Figure 5.5.

For the repeated measures ANOVA of the normalized error area of the mean path Mauchly's Test of Sphericity indicated that the sphericity assumption had been violated, $\chi^2(2) = 8.9$, $p = 0.012$, so the Huynh-Feldt correction, $\epsilon = 0.91$, was applied. There was no main effect of session, $F(1.81, 50.8) = 0.37$, $p = 0.673$. Analysis of the between-subject factors revealed the null hypothesis could be rejected for strength $F(1, 28) = 5.10$, $p = 0.032$, but could not be rejected for shape, $F(1, 28) = 2.32$, $p = 0.139$.

For the repeated measures ANOVA of the normalized step height of the mean path, Mauchly's Test of Sphericity indicated that the sphericity assumption had been violated, $\chi^2(2) = 15.6$, $p < 0.001$, so the Greenhouse-Geisser correction, $\epsilon = 0.70$, was applied. There was no main effect of session, $F(1.39, 38.9) = 2.72$, $p = 0.095$. Analysis of the between-subject factors revealed the null hypothesis could not be rejected for strength, $F(1, 28) = 0.03$, $p = 0.867$, or shape, $F(1, 28) = 0.22$, $p = 0.641$.

5.6 Discussion

All groups increased their step height as a result of training, and this did not significantly degrade during the 26 minutes of post-test. This change was not significantly affected by the strength or shape of the field. This indicates that subjects are able to adapt to the coarse shape of the target for all groups and any effect of shape or strength had a smaller effect size than detectable by this study.

Table 5.3: Damping Parameter Study - Statistics Table for Pooled Post-Test Sessions

Group	Normalized Step Height	t(7)	Corrected p
Lin Low	1.31± 0.27	3.21	**0.030**
Lin High	1.27± 0.25	3.15	**0.016**
Par Low	1.31± 0.19	3.40	**0.035**
Par High	1.29± 0.18	4.52	**0.011**
Group	Normalized Error Area	t(7)	Corrected p
Lin Low	1.01± 0.41	0.08	0.942
Lin High	0.66± 0.20	-4.70	**0.007**
Par Low	0.74± 0.29	-2.52	0.080
Par High	0.63± 0.17	-6.14	**0.002**

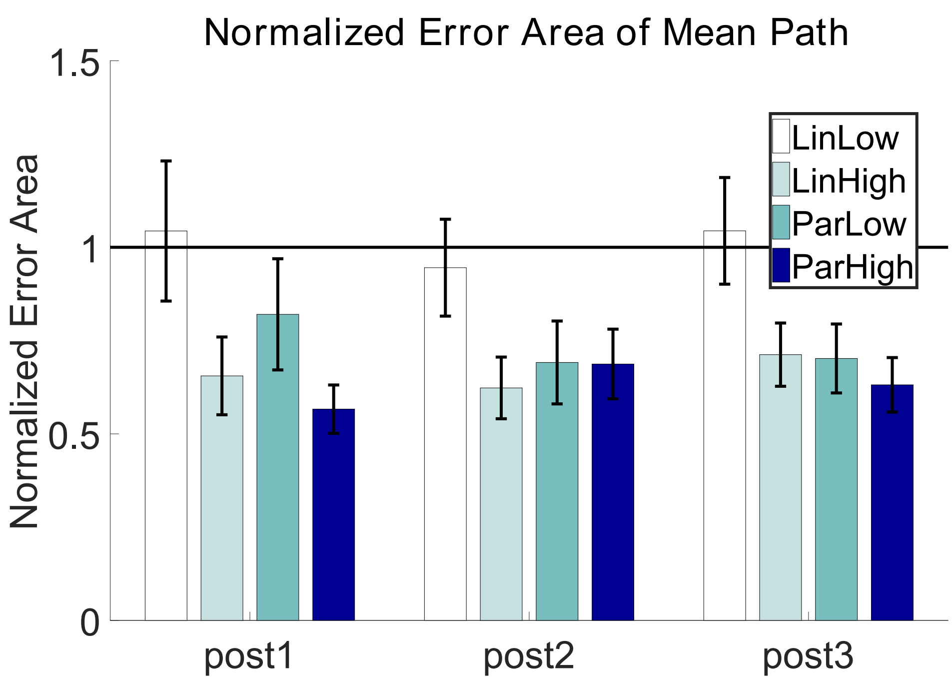


Figure 5.4: Damping Parameter Study - The normalized error area of the post-test evaluations, with the standard error shown.

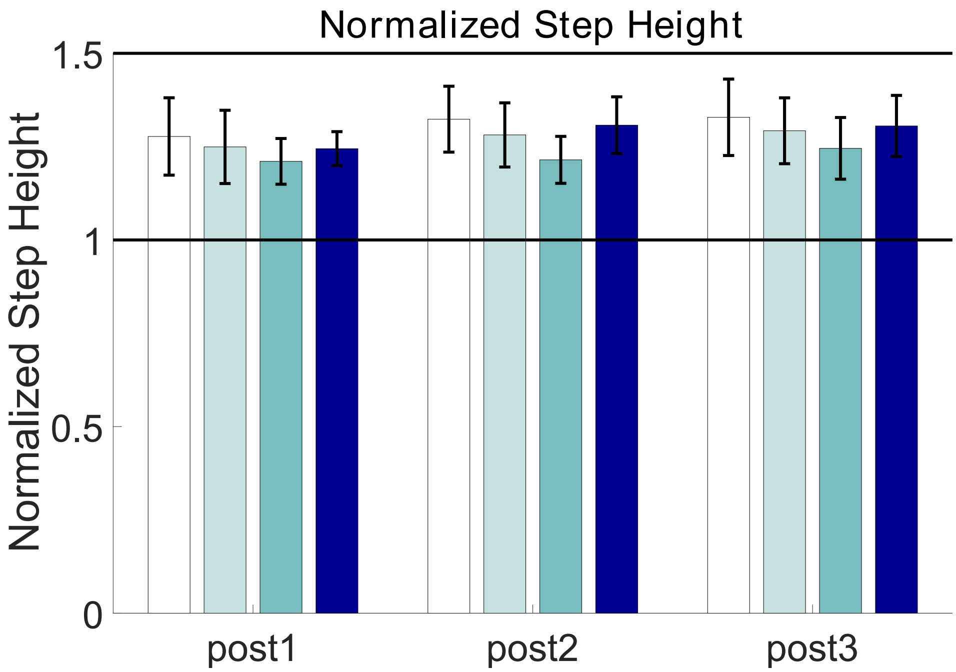


Figure 5.5: Damping Parameter Study - The normalized step height of the post-test evaluations, with the standard error shown.

Only the high strength groups showed adaptation in terms of NEA, which indicates that the faster rate of change of the damping coefficient improved the adaptation to finer details of the foot path. Given this result, it appears that the original hypothesis that a more gradual change in the damping field would allow subjects to follow and retain the path better was not correct. The effect may not be as simple as a faster change in damping coefficient produces better results and there may be an optimal level. Alternatively, the effect may not be a result of the rate change, but could be an effect from decreasing the distance from the path at which the tunnel saturates. With this knowledge, it may be worth exploring the extreme case of saturation at the tunnel width, as would be the case with a step function. This should be explored in both the Cartesian and joint space. If simple step functions are effective, it may indicate that exoskeletons for gait training could be as simple as a rotary damper and a clutch which engages the damper when the subject is outside of the prescribed tunnel. Alternatively, the dampers could be replaced by generators, which would be bypassed while the subject is inside of the tunnel. Rather than generating heat, the energy removed from the system could power the electronics. These, in turn, would only be needed to determine position and engage/disengage the generators, so they would have very little current draw. The batteries could then be smaller and lighter, as they would be frequently recharged, and could potentially be replaced entirely by capacitive storage.

These two open questions, as to the cause of differences seen in the present experiment, could be evaluated in two experiments. First, one could compare the effect of strength using a larger range of values to see if this trend continues with the largest strength being a step function. In this case the damping coefficient increases to the saturation point as soon as it leaves the tunnel. The second experiment could then vary the distance from the path where the saturation point is reached. This could be done by increasing the saturation value so the shape of the field is unchanged until the saturation point. However, this has the problem of the result being affected by the larger possible damping coefficient. Alternatively, the width of the tunnel could be changed, but this leads to the issue of the feedback starting at a greater error distance.

Chapter 6

Comparison of Assistive, Error-Enhancing, and Variable Damping Fields

6.1 Motivation

As discussed in Section 1.3, there is a vast variety of methods for controlling rehabilitative devices. As new control strategies are developed, they are typically evaluated to determine their ability to produce changes relative to an initial measurement from the subject. However, this does not provide an effective way to compare the relative advantages and disadvantages of these control strategies against each other. To do this, formal testing should be done to evaluate them against each other. The following study aims to do that by examining the level of adaptation each produces and the duration this adaptation is retained through post-training evaluations. It addresses rate of adaptation through the use of evaluation in between each training sessions After-effects are evaluated by examining gait changes on the first step after the force is removed. Finally, the amount of force required during training is evaluated to examine the demands on the exoskeleton each strategy produces. This method provides a structure which may be utilized in the future for evaluating various important device features that may be affected by the type of feedback provided.

Table 6.1: Controller Comparison Study - Subjects Information

Group	Sex	Age (yrs)	Mass (kg)	Height (m)
AAN	9 M, 3 F	25.3±3.8	73.8±12.7	1.79±0.09
EE	9 M, 3 F	25.5±5.0	73.3±11.7	1.76±0.09
D	9 M, 3 F	26.1±4.7	72.3±11.5	1.74±0.08

6.2 Protocol

The protocol was similar to that of the damping parameter study; the primary difference was the nature of the haptic force. Thirty-six (N=36) right leg dominant, healthy individuals, with no neurological or physical impairments that would affect their ability to walk in the device or adapt to new gait patterns were recruited, Table 6.1. Each subject performed a single testing session using the ALEX III device. Subjects were separated by gender, and randomly assigned to one of three groups. One group used the assist-as-needed (AAN) force strategy which was described in Section 2.4, another used an error-enhancing (EE) strategy (where the direction of the force used in the AAN force field is reversed, so the force pushes the user's foot away from the path in the direction normal to the path), and the last group used the damping field (D). This was done to compare the effectiveness of these high-level controllers for gait training. All controllers saturated at the same distance from the target path, and the saturation value was set to a value that healthy individuals could tolerate. During post-test evaluations, subject were explicitly asked to continue to follow the path as they had been trained. In the previous study, when subjects were given no specific directions, some assumed they should continue to try to follow the path while others tried to return to their normal gait. There were three main hypotheses for this experiment: 1) the error-enhancing field would produce fewer after-effects in the catch trial than the other two groups, based on previous findings [54]; 2) the damping group would require less force than if it acted normal to the footpath, as the direction of the force interferes with movement and therefore will be more noticeable; 3) the damping group would retain the adaptation longer than the other groups, as humans are thought to naturally try to minimize their energy expenditure while walking, and the damping field capitalizes on this inbuilt optimization.

6.3 Data Analysis

First, one-way ANOVA was performed for height, mass, and age individually with force type as the between-subject factor. This was to confirm that groups were homogeneous. One-sample t-tests were performed on both NEA and NSH by pooling the values for all three post-tests for each subject to evaluate if their mean was significantly different from one, and the Bonferroni-Holm correction was applied. The value of one was used for both tests as it indicates baseline performance. Next, repeated measures ANOVA was performed on the post-test sessions, with session as the within-subject factor, and force type as the between-subject factor. If Mauchly's Test of Sphericity indicated that the sphericity assumption had been violated, the appropriate correction was applied. Post-hoc comparisons were performed using Tukey's HSD. This test was to indicate if there was any degradation of performance over time, and to determine if there was an effect of force type. This repeated measures ANOVA was repeated for the mid-tests to determine if there was a difference in the rate of adaptation. One way ANOVA was performed on the pooled NSH and NEA of the first step without force of the catch trials. This was performed to determine if any group showed an after-effect as a result of removing the force, with Tukey's HSD used for post-hoc comparisons. For each gait cycle during training, the root mean square (RMS) and peak force were found and averaged for each subject. One-way ANOVA was then performed for both force variables with post-hoc comparisons performed using Tukey's HSD. For all tests, $\alpha = 0.05$.

6.4 Results

No significance was found between groups for height ($F(2,33) = 0.925, p = 0.41$), mass ($F(2,33) = 0.049, p = 0.95$), or age ($F(2,33) = 0.090, p = 0.91$). The results of the one sample t-test can be found in Table 6.2. All groups showed statistically significant differences from 1 in NSH and NEA. The normalized error area and normalized step height can be seen in Figure 6.1 and Figure 6.2.

Table 6.2: Controller Comparison Study - Statistics Table for Pooled Post-Test Sessions

Group	Normalized Step Height	t(11)	Corrected p
AAN	1.41±0.26	5.50	**<0.001**
EE	1.40±0.36	3.84	**0.003**
D	1.38±0.27	4.97	**<0.001**
Group	**Normalized Error Area**	**t(11)**	**Corrected p**
AAN	0.67±0.39	-2.87	**0.015**
EE	0.76±0.28	-2.92	**0.028**
D	0.70±0.28	-3.62	**0.012**

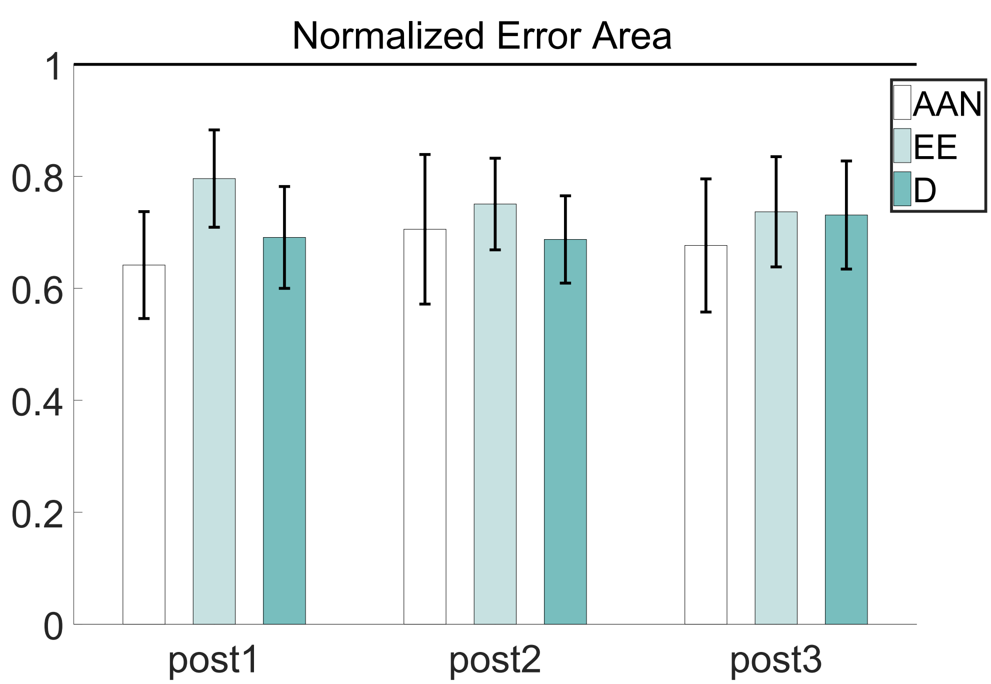


Figure 6.1: The normalized error area of the post-test evaluations for the controller comparison study, with the standard error shown.

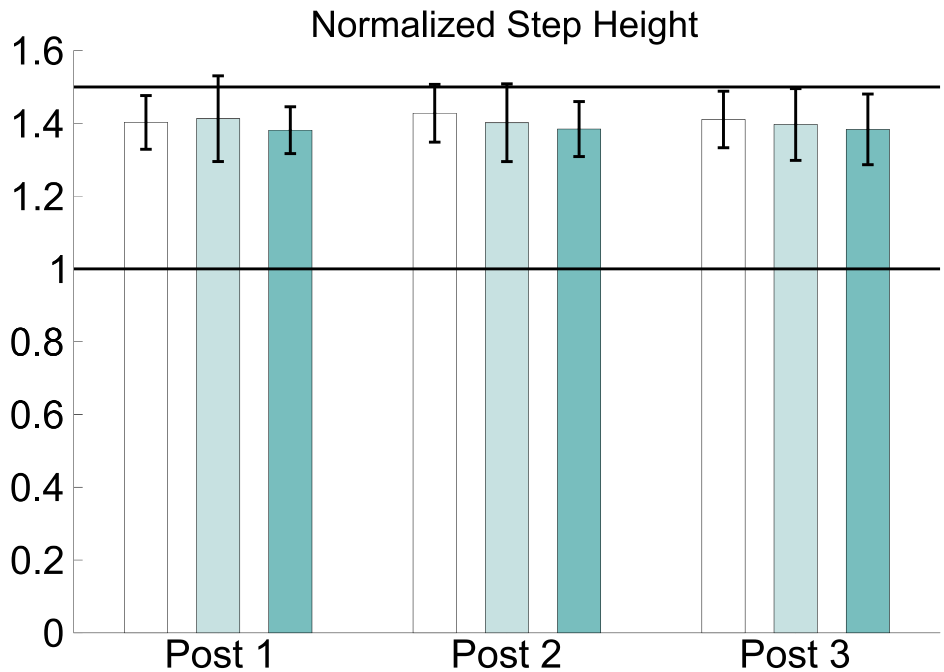


Figure 6.2: The normalized step height of the post-test evaluations for the controller comparison study, with the standard error shown.

For the repeated measures ANOVA of the NSH of the mean path in the post-test evaluation, Mauchly's Test of Sphericity indicated that the sphericity assumption had been violated, $\chi^2(2) = 14.7$, $p = 0.001$, so the Greenhouse-Geisser correction, $\epsilon = 0.73$, was applied. There was no main effect of session, $F(1.46, 48.2) = 0.071$, $p = 0.877$. Analysis of the between-subject factors revealed that the null hypothesis for force type could not be rejected $F(2, 33) = 0.033$, $p = 0.968$.

For the repeated measures ANOVA of the NEA of the mean path in the post-test evaluation, Mauchly's Test of Sphericity indicated that the sphericity assumption had not been violated, $\chi^2(2) = 2.26$, $p = 0.323$. There was no main effect of session, $F(2, 66) = 0.016$, $p = 0.98$. Analysis of the between-subject factors revealed the null hypothesis for force type could not be rejected, $F(2, 33) = 0.22$, $p = 0.802$.

For the repeated measures ANOVA of the NSH of the mean path in the mid-test evaluation, Mauchly's Test of Sphericity indicated that the sphericity assumption had not been violated, $\chi^2(2) = 3.85$, $p = 0.15$. There was no main effect of session, $F(2, 66) = .101$, $p = 0.904$. Analysis of the between-subject factors revealed the null hypothesis for force type could not be rejected $F(2, 33) = 0.56$, $p = 0.58$.

Table 6.3: Controller Comparison Study - Table for NEA and NSH during mid-tests

	NSH		
Group	Mid-Test 1	Mid-Test 2	Mid-Test 3
AAN	1.37 ± 0.34	1.27 ± 0.28	1.36 ± 0.39
EE	1.41 ± 0.24	1.48 ± 0.25	1.47 ± 0.41
D	1.39 ± 0.25	1.40 ± 0.32	1.38 ± 0.30
	NEA		
Group	Mid-Test 1	Mid-Test 2	Mid-Test 3
AAN	0.85 ± 0.32	0.73 ± 0.30	0.79 ± 0.38
EE	0.74 ± 0.19	0.69 ± 0.32	0.80 ± 0.40
D	0.67 ± 0.72	0.79 ± 0.46	0.82 ± 0.57

For the repeated measures ANOVA of the NEA of the mean path in the mid-test evaluation, Mauchly's Test of Sphericity indicated that the sphericity assumption had not been violated, $\chi^2(2) = 0.79$, $p = 0.68$. There was no main effect of session, $F(2, 66) = 0.716$, $p = 0.49$. Analysis of the between-subject factors revealed the null hypothesis for force type could not be rejected, $F(2, 33) = 0.041$, $p = 0.96$.

For the one-way ANOVA of the pooled NSH in the catch-trials, the null hypothesis could not be rejected, $F(2, 33) = 3.28$, $p = 0.05$. This was also the case for NEA, $F(2, 33) = 1.13$, $p = 0.34$.

For the one-way ANOVA of the peak force during training, Figure 6.3, the null hypothesis could be rejected, $F(2, 33) = 49.8$, $p < 0.0005$. This was true for NEA as well, Figure 6.4, $F(2, 33) = 19.8$, $p < .0005$. Tukey's post-hoc test revealed that the damping field used significantly less force (peak and RMS) than both of the other two groups ($p < 0.0005$ for all). There was no statistically significant difference in the force applied between the assist-as-needed and error-enhancing groups ($p = 0.99$ for peak and $p = 0.64$ for RMS).

Table 6.4: Controller Comparison Study - Table for Pooled NEA and NSH during the First Step in the Catch Trials

Group	NSH	NEA
AAN	1.34 ± 0.11	0.92 ± 0.31
EE	1.39 ± 0.24	0.98 ± 0.27
D	1.60 ± 0.36	1.12 ± 0.40

Table 6.5: Controller Comparison Study - Table for Pooled Training Force

Group	Peak Force	RMS Force
AAN	23.3±8.0	8.34±5.12
EE	23.7±5.9	9.69±3.58
D	3.1±1.6	1.04±0.51

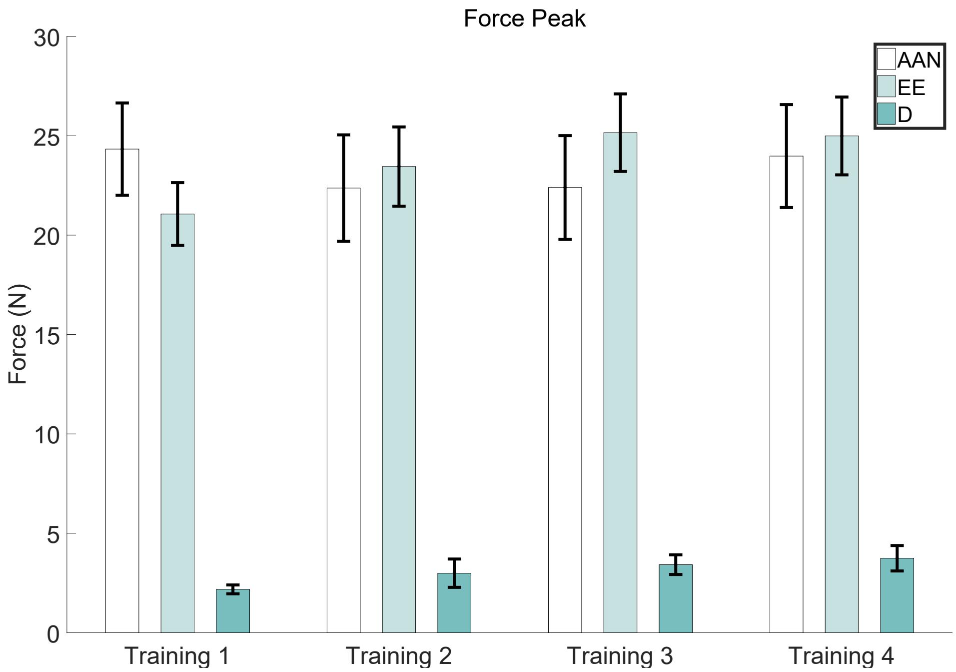


Figure 6.3: The average peak force during training for the controller comparison study, with the standard error shown.

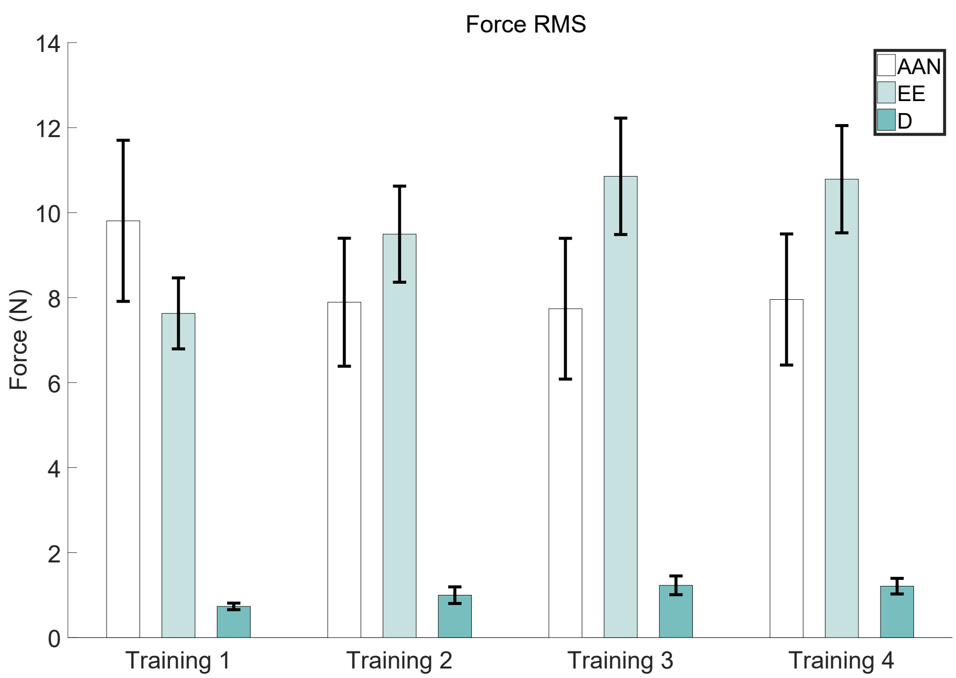


Figure 6.4: The average RMS force during training for the controller comparison study, with the standard error shown.

6.5 Discussion

All groups were able to show adaptation in both NEA and NSH in the post-tests; however, there was no statistically significant difference between the groups in the level of adaptation, the speed of adaptation, or the after-effects when the force was suddenly removed. This last finding does not support our first hypothesis that the error-enhancing field would produce fewer after-effects. From prior findings, it was expected that the AAN strategy would produce greater after-effects than the EE strategy [54]. This may be due to the nature of the device, as the motors are not collocated with the joint and there is some compliance in the belts used to transmit the motor torque. As a result of this compliance, the device is tuned to be slightly damped to prevent oscillations in the leg. This may mask some of the after-effects resulting from the removal of the AAN force.

As all groups retained their adaptations till 26 minutes following training, it is unclear if the damping field actually produced longer lasting adaptations. This finding cannot validate our third hypothesis that the damping group would retain the adaptation longer than the other two groups, as during the testing period there was not a statistically significant change in the performance of any group. This result, while unexpected, would indicate that for neurally intact individuals the type of force applied to the leg may play a less important role, and that as long as the wearer is engaged and challenged the outcomes may be similar. This may be why in the previous study, the linear low strength group had lower performance, as the linear low strength damping field did not provide sufficient challenge, by interfering with the subjects ability to move their foot. This may be a better way of exploring what force types are best suited for impaired populations. Instead of trying to find the single best type for an individual population it may be better to use any that can be tolerated by the patients while still providing a challenging task. This could be accomplished through testing each type of force for every patient and using a mixture of the ones that are challenging, but achievable. This can be more advantageous as well, as it can encourage the use of various types of forces to produce variable practice, which has extensively demonstrated improvements in retention and performance [79–83].

The most useful finding of this study is that damping can substitute AAN or EE forces, providing similar levels of adaptation for at least the measured period of time, and can do this using 1/7 of the force, which confirms our second hypothesis that the damping field would require less force than the other two groups.

This may be a paradigm changing finding for providing therapy to individuals with sufficient functional ability. Because the amount of force required for this type of therapy is so much smaller than that of the other strategies, the size of devices can be greatly reduced as the stress on parts will be lower. The force generating components can be smaller, and use less power or even potentially generate power as the forces are dissipative. As a result of the reduced power requirements, mobile, wearable systems become more realistic. Batteries can be much smaller, or non-existent with the use of capacitive storage, as power would only need to be supplied to the control system if regenerative breaking is used to apply the damping force. The topology of the exoskeleton can also be changed; Because the nature of the force is dissipative and small it may lend itself well to spooled cable reels that can be connected to the leg by routing through fabrics. A sufficiently compact and light design would allow patients to wear it all day under their normal clothes with the addition of a hip pack to house the electronics and cable reels. This would allow the patient to receive therapy at anytime of day that suits them, while allowing them to select when they would like to train and when they wouldn't. This would allow them to get much greater levels of repetitions as they can then do therapy at any time of day without needing to modify their daily schedule. There are challenges with this in terms of creating a comfortable and unintrusive device which patients would be willing to wear. However, it is worth exploring, as the amount of practice is related to the amount of improvement seen [84] and compliance with at home therapy can be low, unless it becomes part of the patient's daily routine. As visual feedback is challenging in a mobile setting, auditory feedback can potentially be added to enhance the therapy [29].

Several pilot studies have found that the adaptations in healthy individuals do not transfer overground and this has been supported in the literature [54]. The use of mobile training devices may resolve this issue. Additionally, the use of auditory feedback could be used to wean subjects off of haptic feedback to produce overground transfer.

Chapter 7

Design Choices and Iterations

7.1 ALEX II

There were several objectives in the design of ALEX II which were intended to overcome some of the limitations of ALEX I. The first was the addition of anterior/posterior motion of the pelvis. When subjects walked in ALEX I, many noted that they were prevented from performing anterior motion with their pelvis, which felt unnatural. This is due to the nature of walking which can be modeled as an inverted pendulum when the person is in the single support phase. As the person moves through their gait, the pelvis traces an arc which requires both superior/inferior and anterior/posterior motion. Secondly, ALEX II should be used on either leg. This is because stroke can affect either side of the body, so restricting the device to just the right leg limits the population which can use the device. Third, the range-of-motion of the joints should be independent of the segment length. In ALEX I, the motion was provided by linear actuators. As the limb length changed, the available stroke length of the motor changed, because it spans the adjustment by running parallel to the segment, Figure 7.1. As a result, shorter subjects were not able to fully flex their knee as they would reach the limit of the motor stroke. Additionally, we wanted the adjustment and donning procedure to be as simple as possible. In ALEX I, a shoe insert was attached to the device to measure the ankle angle, Figure 7.1. To don the device, a subject's shoe must first have this insert placed into it. Then the subject must try to place their foot into the shoe and insert while putting themselves in the rest of the device. This procedure is difficult, especially when working with individuals with motor control issues.

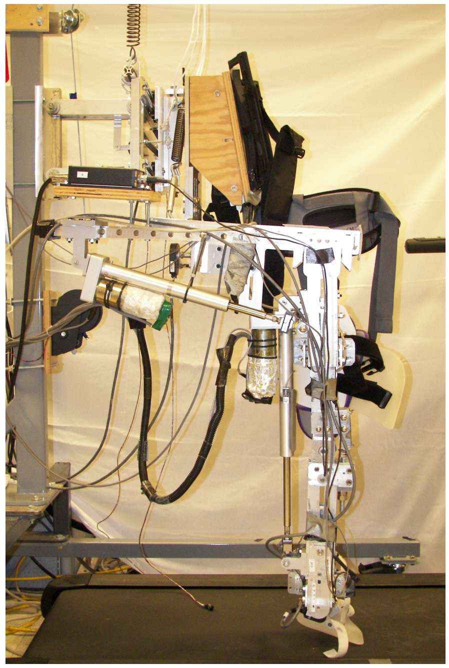


Figure 7.1: ALEX I

7.1.1 The Support System

To achieve anterior/posterior motion in the support system, initially a series of prismatic joints was used, Figure 7.2. This method resulted in very limited functionality, as the gravitational load was supported by a simple spring. As a result, the force applied to the subject would change during the gait cycle as the subject moved above and below the equilibrium position of the spring. Additionally, a large moment was created on the sliders, which relied on low friction sliding surfaces, resulting in the system binding. Due to these issues, this concept was abandoned in favor of the design presented in Chapter 2. By moving to links supported by revolute joints, gravity balancing could now be used, which removed the relationship between position and the force felt at the pelvis. Additionally, when early testing was performed, it was determined that the position of the pelvic attachment should be adjustable in the anterior/posterior direction, to account for different body sizes. To achieve this, a scissor jack mechanism was added, Figure 7.3.

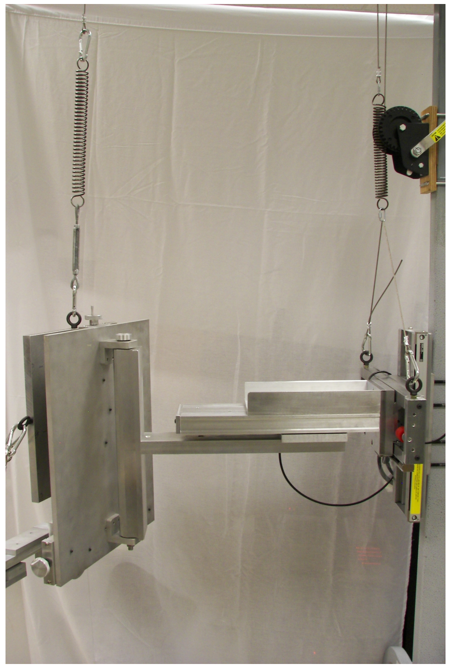


Figure 7.2: The initial support system for ALEX II, consisting of a series of prismatic joints.

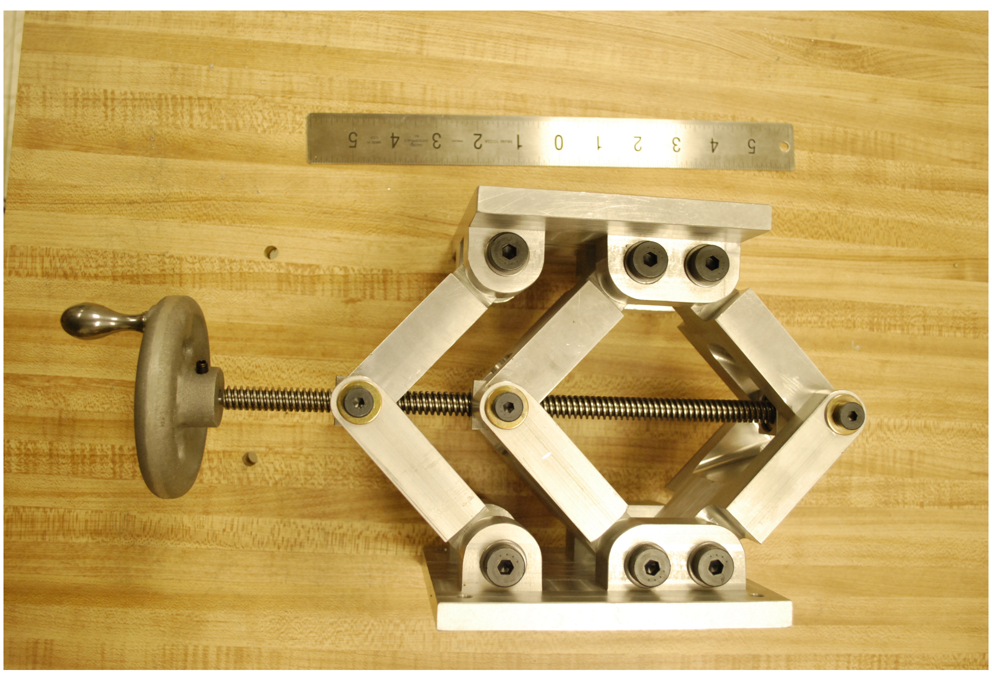


Figure 7.3: The scissor jack used for adjusting the pelvic attachment point.

7.1.2 The Leg

To realize the design goals of the new system, several features were initially added to the leg of ALEX II. First, revolute motors were used in place of linear actuators to remove the range-of-motion dependence on segment lengths. Second, the leg can be rotated 180° around the hip abduction axis to switch the device to the other leg. Lastly, the ankle measurement system can be placed on the subject before getting in the device. There were several iterations, as discussed in the next paragraph, before reaching the final design.

The ankle joint initially consisted of a shoe insert which could be placed in the shoe prior to the subject donning the device. When the subject stepped into the device, the shoe insert would clip into a receptacle on the exoskeleton. However, this approach added a significant amount of weight at the distal portion of the leg. Additionally, the design as implemented had its axis of rotation parallel to the hip and knee flexion/extension axes. This is not the typical axis of rotation of the ankle, and as a result, it interfered with walking. Due to these issues, the design was changed to a lightweight polypropylene goniometer that attaches over the shoe before the subject dons the device.

Originally, an additional degree-of-freedom was added to the hip joint to provide three rotational degrees-of-freedom. This additional degree-of-freedom allowed medial/lateral rotation of the leg. Due to the leg and cuff being near cylindrical shapes, the human leg was not firmly constrained to the device about this axis. During initial testing, it was found that the robotic leg would rotate about this axis during gait due to the inertia of the leg, while the human leg would not. To prevent this unintended motion, the medial/lateral rotation was locked in the first design and removed completely from later designs.

7.2 ALEX III

The primary objective of the ALEX III was to provide bilateral gait training. Additionally, it was desired that the system be fully active to allow for complete control of the device and to prevent parasitic motions that result from the passive dynamics of the device. It was also desirable to lower the inertia of the leg to improve transparency, and to move as much of the device behind the user as possible in order to provide

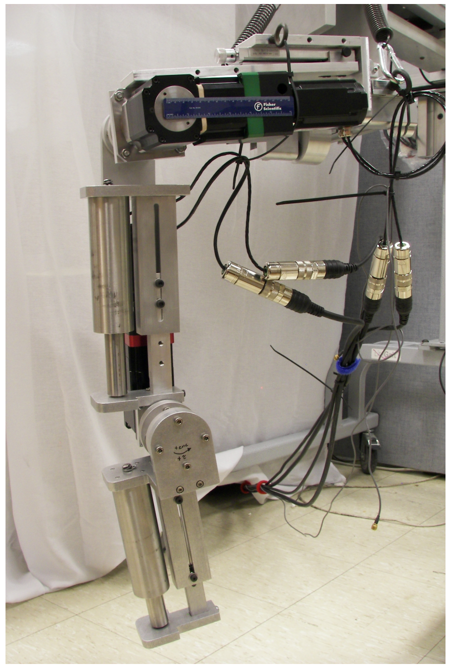


Figure 7.4: The initial leg length adjustment for ALEX II, which consisted of a screw and nut for adjusting the length which could then be locked using screws and hex keys.

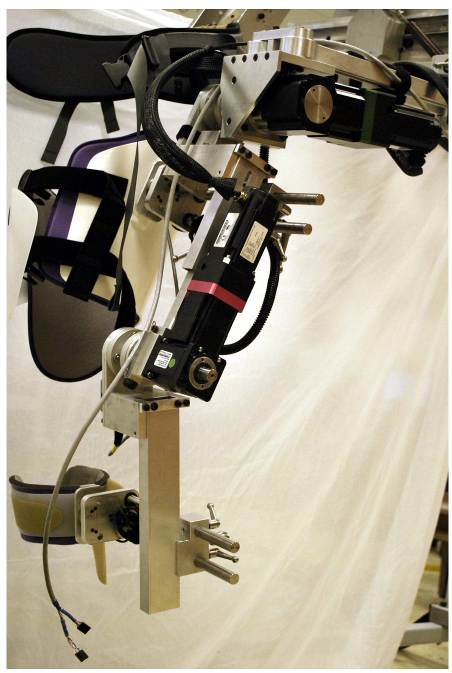


Figure 7.5: In the updated design the length adjustments are performed using a screw with a handle in a slot. The thigh and shank components are now also connected to the proximal portion of the leg through an L-beam instead of a flat plate.

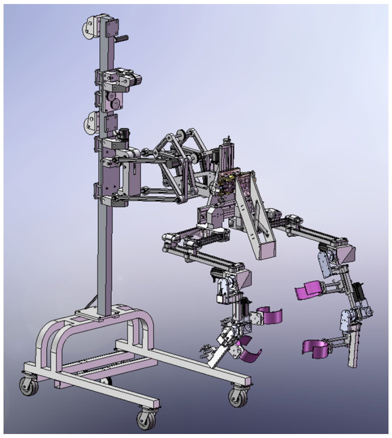
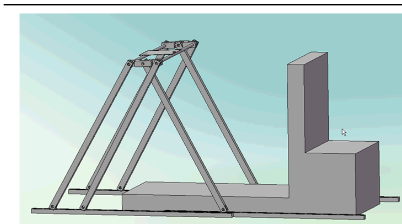
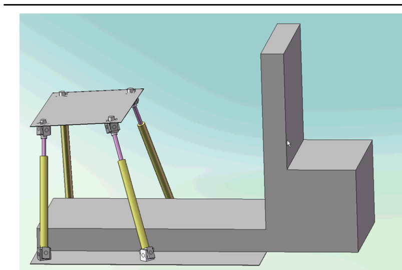
a more open feeling in the device, as well as to allow free arm swing. Finally, each degree-of-freedom should be able to provide sufficient torque to complete a typical gait cycle, given a passive subject. This also requires considering the dynamics of the system and the motor velocity.

7.2.1 The Support System

Several different concepts for the support system were explored before settling on the final design, Table 7.1. The first concept was to simply use the support system for ALEX II. This idea however did not meet the criteria for active control. It could be modified by placing motors on each degree-of-freedom, but these motors would interfere with the linkages due to space constraints. The next concept was similar to what was chosen for the final design. It consisted of two PRRRP manipulators, providing superior/inferior and anterior/posterior motion, attached to each other at the top to provide a location for lateral motion and rotation about the vertical axis. For this design, both sides would need to be driven in an identical manner and lateral forces at the top would create large moments on the members. Another design was motivated from Stewart platforms. However, with this design each motor would need to continuously support a portion of the gravitational load, and analytical solutions are not possible unless special geometries are chosen. The final concept was a serial arm. This could be commercially purchased, but would be prohibitively expensive. It could also be made in house, but because the load would be supported at the end of the arm, there would be large moments generated at the more proximal motors. This factor made the closed chain mechanisms more attractive in the design, and the ability to analytically solve the PRRRP manipulator made it the most attractive design choice.

To address the limitations of the simple PRRRP manipulator, several design changes were made. As discussed in Chapter 4, universal joints were added to the ends of the legs, in addition to angling the legs. This prevents the large moments that would be generated in a simple PRRRP manipulator. In addition to those desicions, there were two more choices to make for the actuation of the support system. The first concerned the source of the driven prismatic joints. When evaluating if these should be purchased or produced in house, it was found that the cost of one commercially available component would be greater than the cost of producing all of them in house. The second concerned how they would be driven. If each prismatic joint had its own motor, force control could result in the sides being driven to different positions

Table 7.1: Conceptual Designs for ALEX III Support System

Support System Concept	Pros	Cons
 Bilateral ALEX II	• Quick to implement as it is already designed • Limitations are known	• Passive • Large size • Heavy, with noticeable inertia
 PRRRP Manipulator	• Few moving parts • Large motors are stationary	• Both sides must be driven identically • Lateral forces must be stabilized
 Stewart Platform Motivated Manipulator	• Large motors are on the ground • Uses few parts other than the actuators	• Requires special geometry • Multiple large actuators are needed to support load
Serial Arm Manipulator	• Commercially available • Quick to implement • Large workspace	• Expensive • Large motors are required to support a load at the end of the arm

if the loads were not completely equally distributed. This would result in an unintentional rotation of the device. If appropriately controlled this could be an additional feature, but it was determined that it would increase the complexity of the design more than necessary and could be added later if there were a specific reason to do so. Due to these considerations, the prismatic joints were designed in house so that the front pair was driven by one motor and the back pair was driven by a separate motor.

The legs of the support platform were designed for use with a split belt treadmill, to allow for normal gait in individuals with thigh and shank lengths from 35 cm to 45 cm. At the time it was unclear if the device would be used with a split belt treadmill, so the legs were designed to be interchangeable with smaller legs that could be used with a traditional treadmill. However, in the current implementation the longer legs are being used with a traditional treadmill which is placed on an elevated platform. This setup provides a convenient location to route cables and does not require the support legs to be switched.

One final design modification to the legs was made after the support system was tested. It was found that when fully assembled, there was a small amount of motion in axis of rotation of the bearings. This small, angular motion at the ends of the leg resulted in a significant amount of lateral motion at the top of the support system. To prevent this motion, tensioned cables were placed across the two rear legs. This addition constrains them, preventing the unwanted motion.

7.2.2 The Legs

The primary design objective for the leg was to reduce interference with gait. This was achieved by focusing on improving the transparency of the legs and minimizing secondary interactions with the user, such as the arms hitting the device when they swing. These objectives were interrelated and design choices which achieve one aid the the achievement of the other. The first design choice was to use non-collocated motors. Because the motors are non-collocated, they no longer are required to have their axis of rotation in-line with the human joint axis, so there can be less material on the sides of the user. This allows the user to freely swing their arms without making contact with the robot. Additionally, by moving the motors off the leg, the overall inertia of the leg can be reduced. This has the benefit of improving transparency, as there is

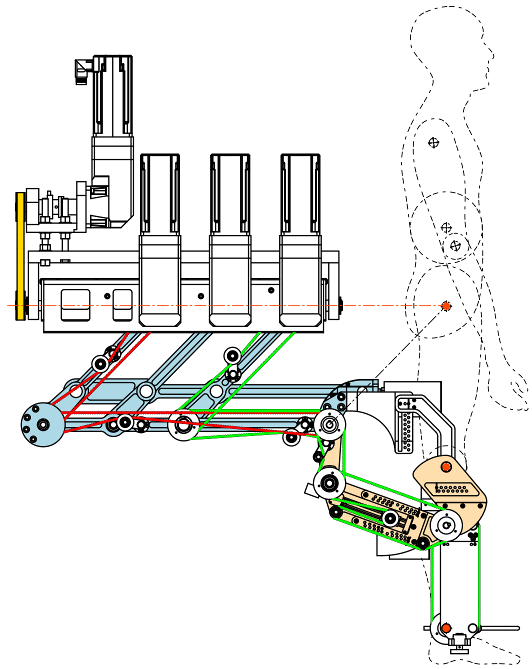


Figure 7.6: Line drawing of the leg with the knee belts in red and the ankle belts in green. The hip rigid links are light blue and the knee links are light orange.

less inertia that must be compensated for through the controller. However, moving the motors off the leg increases the complexity of the leg, as there must be a method of transmitting the torque from the motor to the leg.

Several transmission methods were considered for the leg. The first was to use rigid links. This creates the least compliant design of all of the transmissions considered. Also, as the segment lengths change, the transmission ratio for fixed length rigid links also changes. This can be accounted for by making the rigid links adjustable. This method also adds mass to the system, as the rigid links are more massive than the other two methods considered. The second method considered was a belt driven system. This has the advantage of being lightweight and having components that are commercially available. They have a greater degree of compliance, which may affect the performance of the system. With any method there will need to be some form of adjustability for the subject size. When spanning a link which is adjustable they will also require some form of adjustment. A third method, drum capstans, have a higher stiffness than belts so the system response may be better, but they still face many of the challenges that exist with belts, in addition to requiring large drums to match the bend radius of the cables. Considering all of these factors, the final

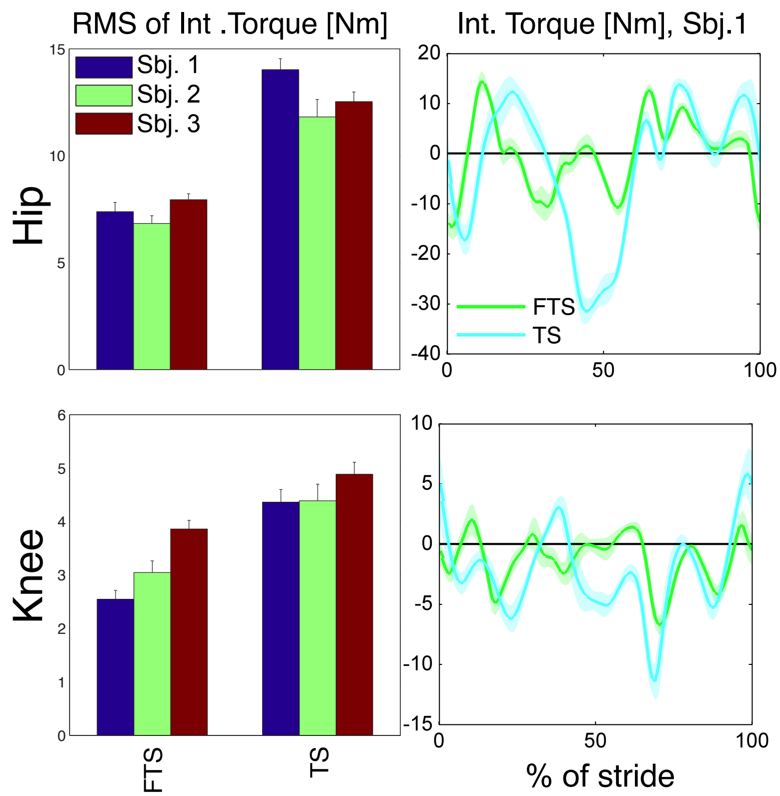


Figure 7.7: RMS interaction torques and average interaction torques over the gait cycle for a representative subject (shaded areas and error bars represent $\pm$SD).

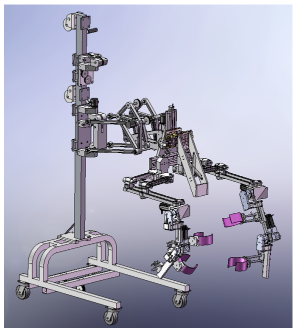
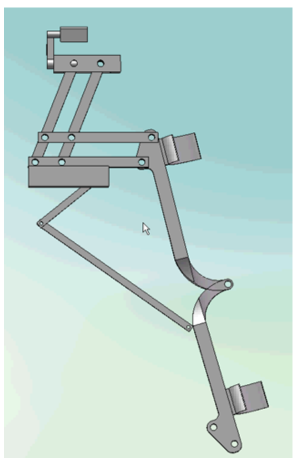
solution uses a combination of both rigid links and belts. The hip is driven solely through rigid links. The knee uses rigid links, but power is transmitted to this linkage through belts which run over the hip's rigid links, Figure 7.6. While the lengths of the hip's rigid links are fixed, the links for the knee are adjusted with the limb length to maintain a 1:1 transmission ratio from the motor gearbox output to the knee joint. The transmission for the ankle is significantly more complex, as it must traverse the rest of the leg. Its belts run over the hip links in much the same way as the knee belts. Belts then run across the knee links which requires a tensioning mechanism so the belts can accommodate changes in length of the links. Finally, the transmission must cross the length of the shank before reaching the ankle. This requires the same degree of adjustability that was needed to accommodate changes in length of the knee links.

In addition to lowering the inertia of the leg, force/torque sensors were placed at the human/robot interaction points. These were used for feedback for the force control. This means that the inertial load and friction in the system that would normally be transmitted to the user is measured as an error term and is subsequently accounted for by the control system. This improves the transparency without requiring a model of the system which would depend on noisy acceleration and velocity values that result from differentiating the measured position of the leg. The use of these models in a feed-forward term may still further improve transparency, but these models are not currently implemented. This design choice was first validated in ALEX II, by comparing the use of torque sensors at the motor output, as traditionally used, against the use of force/torque sensors at the user interface before this design choice was made [29].

This validation was performed by testing three individuals who walked in the device while it was controlled to provide no interaction force, but to compensate for gravitational loads and friction. Each subject walked in the device at 2.4 mph, while the feedback loop was being closed on the force/torque sensor (FTS). They then repeated the trial while a torque sensor at the motor's gearbox output (TS) was used for feedback.

Figure 7.7, left side, shows the average RMS value of τ_{int} over sessions FTS and TS: the FTS controller effectively reduced the RMS of the hip interaction torque and, to a lesser extent, also the knee interaction torque (τ_H -42.1%, τ_K -31.1% on average). Torque profiles over the gait cycle (GC) show that peak values were reduced even further (Fig. 7.7, right side). The less evident effects measured at the knee joint can be explained by the fact that the moment of inertia of the shank link about the knee joint was smaller than the overall moment of inertia of the robotic leg reflected at the hip joint. These reductions in interaction torques indicated that this method is effective at reducing the interaction force and that this method would be desirable for use in ALEX III.

Table 7.2: Conceptual Designs for ALEX III Legs

Leg Concept	Pros	Cons
 Bilateral ALEX II	• Quick to implement as it is already designed • Limitations are known	• Large size • Heavy, with noticeable inertia
 Rigid Linkage Driven	• Free arm movement • Motors proximally located • Device out of sight • Nothing to tension	• Transmission changes with limb length • Aligning a virtual hip joint is more difficult
Belt Driven System	• Motors can be mounted even more proximally • Light weight	• Requires tensioning • Length changes with changes in limb length • Adds compliance into the system
Drum Capstan Driven System	• Motors can be mounted even more proximally • Lightweight	• Requires tensioning • Length changes with changes in limb length • Needs large drum

Chapter 8

Contributions and Future Work

8.1 Contributions of the Current Work

8.1.1 Development of the ALEX II

The creation of ALEX II solves some of the limitations of the previous version of the exoskeleton, providing a highly versatile platform for training individuals with stroke, as well as improving our understanding of RAGT. In addition to the studies presented above, ALEX II has been used to study gait training in stroke survivors as well as healthy young adults. It has been used to retrain chronic stroke survivors to a gait pattern based on an unimpaired size and age matched individual, and found that improvements in over-ground walking speed persisted up to 6 months following training [85]. In healthy young adults, there has been extensive use of ALEX II to study how to enhance RAGT and to improve our understanding of what occurs during training. In one study, while healthy young adult subjects trained, we recorded EEG data to understand what areas of the brain are engaged during training [86]. The device has also been used to explore the ability of auditory feedback to replace visual feedback, which may have advantages for populations who have trouble interpreting visual information due to impairment [87, 88]. Additionally, it has been used to compare the effect of different types of haptic feedback, similar to the controller comparison study [54]. Alex II has also been used to explore how the design changes the performance of exoskeletal devices. In the first of these studies, ALEX II was used to show that placing force sensors at the attachment points to the user, as opposed to placing them at the motor output, improves the transparency of the device [29]. The second of these addressed the fact that there is always some degree of misalignment between the human and robotic joints, and characterized the effect of this misalignment on the forces applied to the user [89]. Finally, in an effort to understand how external devices can reduce muscle activity during walking, the device was configured to apply assistive forces at hip during specific phases of gait. To simplify the robot for this application, the lower portion of the leg was removed [90]. There are currently plans to use ALEX II to evaluate the use of the damping field to retrain stroke survivors.

8.1.2 Development of the ALEX III

While the ALEX III device has been used less extensively, its development has allowed the exploration of new ways of providing therapy, and has an increased potential to apply therapy to non-hemiparetic populations. The first set of studies explored the use of one leg to create the target path for the other leg. Traditionally for stroke therapy, gait templates based on unimpaired control subjects have been used. While this will create a more natural gait, it still may not be ideal for that particular person. An alternative would be to train them to their own gait pattern. However, without a template from before their incident, it is not possible to know how they walked before the impairment. ALEX III has the ability to create a target path based on the unimpaired leg, which adapts to changes in gait as the person walks. This allows the person to train to their own gait pattern, and as their impaired leg improves, hopefully fewer compensation strategies will be used by their unaffected leg, creating a more natural and symmetric gait pattern. The purpose of that first study was to validated this strategy in healthy young adults with a weighted leg [44]. In the future, the device will be used to evaluate different types of visual feedback for bilateral training, as well as the effect of pelvic perturbations on training, and the use of the ankle joint in therapy. We also intend to extend its use to new populations, such as spinal cord injury and cerebral palsy.

8.1.3 Pelvic Freedom's Effect on Adaptation

An understanding of how the design of a system affects its performance is fundamental to any design problem. Up to now, systematic study of exoskeleton design has been lacking. This work takes the first step to understand how the amount of freedom allowed at the pelvis affects the primary goal of the system, which is to create gait adaptations. It was found that allowing a large amount of pelvic freedom or greatly restricting the pelvic freedom improves the level of adaptation in healthy young adults. Restrictions of the pelvic motion have a greater impact on adaptation than the restriction of minor leg degrees-of-freedom such as hip ab/adduction. These findings, if validated in impaired individuals, can guide the design of future exoskeletons and will hopefully improve the quality of physical therapy by providing devices that have only the necessary components for therapy without additional complications and costs.

8.1.4 Damping Based Training Devices

This work created a novel high-level controller designed to incorporate the motor learning ideas of variable practice, banded feedback, and descriptive knowledge of performance. This high-level controller provides increased damping force as the error increases beyond a specified threshold. It was experimentally validated to be effective in creating gait adaptations in healthy young adults, and it was found that for the tested values, a faster rate of change of the damping coefficient produced adaptations that more closely followed the trained path. This control strategy was then compared to previously used strategies, and while there was no difference in terms of adaptation levels, the damping field used a much smaller amount of force to achieve the same result. This finding may influence the design of exoskeletons in the future, as exoskeletons requiring less force can be made lighter, more portable, and potentially less expensive, so that they can be used in daily life and increase the amount of therapy patients can receive.

8.2 Suggestions for Future Work

While the findings in healthy young adults are promising, impaired populations may not respond in the same way. As a result, all of the above work needs to be replicated in clinical populations, as they will have different functional abilities and may react to training paradigms differently than unimpaired individuals. This work was originally targeted at stroke, but there are applications in cerebral palsy and spinal cord injury, among other conditions. These replications should not only look at which of these methods are most effective for a specific condition but should explore how they can be combined to improve the overall outcome, and the appropriateness of the therapy depending on the severity of the condition.

To gain more insight into how the pelvic motion affects training, the DOF work can be extended to look not only at the restriction of the pelvis, but also at training new gait patterns in the presence of perturbations, which would increase the level of challenge of the task and create a gait pattern which may be more robust to real world situations. Robust real world transfer is still challenging, but pelvic restrictions may make this transfer more challenging. It should also be explored how the pelvic freedom affects overground transfer in impaired populations and how pelvic freedom can be staged to increase the rate of improvement. We hypothesize that early therapy or therapy in severely impaired populations may benefit from reduced pelvic freedom, as it will be less challenging and more easily tolerated. As they improve, they will benefit from the increased balance challenge created by increased pelvic freedom.

The damping field is in an extremely early stage in its development, and there are many open areas as a result. First, does the result that a faster rate of change of damping coefficient produces better adaptation continue to the extreme case of a step function? Secondly, do the findings apply in the same way if the control is done in the joint space? If these are both the case, mechanical design as well as control of these devices can be much simpler because simple clutched dampers can be used without the need for force feedback. Once these are known, we can address what the best design topologies are for achieving this damping field. There are several possibilities, each with the benefits and drawbacks. The clutched damper mentioned before could be used, or variable dampers. Friction breaks could also be used, which eliminates the need for fluids and thereby the potential for leaks, but would need more frequent replacement of pads as they wear. Finally, regenerative breaking could be used to produce the energy required to power the exoskeleton. Any of these methods could be combined with either rigid elements or lower profile cables to produce a wearable device. The design of this new type of exoskeleton needs to be performed and then its ability to produce improved outcomes in impaired populations needs to be validated. With these new designs, the exoskeletons may be inexpensive enough as well as light enough for everyday use, and this could greatly increase the amount of therapy which can be provided. Additionally, if it is able to be used overground for training, this may increase the likelihood of overground transfer.

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