Proceedings of 2011 International Conference on
Rehabilitation Robotics
June 29 – July 1, 2011, Zurich, Switzerland

Design and Evaluation of Mina

a Robotic Orthosis for Paraplegics

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Abstract— **Mobility options for persons suffering from paraplegia or paraparesis are limited to mainly wheeled devices. There are significant health, psychological, and social consequences related to being confined to a wheelchair. We present the Mina, a robotic orthosis for assisting mobility, which offers a legged mobility option for these persons.**

Mina is an overground robotic device that is worn on the back and around the legs to provide mobility assistance for people suffering from paraplegia or paraparesis. Mina uses compliant actuation to power the hip and knee joints. For paralyzed users, balance is provided with the assistance of forearm crutches. This paper presents the evaluation of Mina with two paraplegics (SCI ASIA-A). We confirmed that with a few hours of training and practice, Mina is currently able to provide paraplegics walking mobility at speeds of up to 0.20 m/s. We further confirmed that using Mina is not physically taxing and requires little cognitive effort, allowing the user to converse and maintain eye contact while walking. [1]

Keywords-component; Powered Orthosis, Exoskeleton, Paraplegic, Gait generation, Locomotion, Orthoses, Robotics, Mobility device.

I. Introduction

Mobility options for those suffering from paraplegia or paraparesis are limited. The most common means of locomotion utilizes the wheel and the required infrastructure (ramps, roads, smooth surfaces, etc.) and 69.8% of spinal cord injured (SCI) paraplegics use a manual wheelchair as their primary means of locomotion [1].

A person in a wheelchair can only access a small fraction of the locations that a pedestrian can access. Wheelchairs have trouble on curbs, stairs, irregular terrain such as hiking trails, and narrow corridors. Even with advances in powered wheelchairs, such as the iBot (http://www.ibotnow.com/), mobility is still limited to relatively smooth terrain, which excludes much of the natural outdoors. Additionally, being confined to a wheelchair has significant consequences on health and quality of life, as well as psychological and social implications. Health related issues include pressure sores, circulation, and bone density loss and changes in fat mass [2,3,4].

Applying robotics to lower extremity orthosis design offers new mobility options for those limited to a wheelchair. There are a few devices that are in development that have started this transformation.

The first is the ReWalk from Argo Medical Technologies (http://www.argomedtec.com/). This device is worn around the back and legs. Electric motors at the hips and knees move the legs to provide locomotion and the user provides balance with the aid of forearm crutches. The device is controlled by torso motions and a push-button interface.

Another device that was recently introduced is the eLegs from Berkeley Bionics (http://berkeleybionics.com/). Similar to the ReWalk, this device attaches to the user's back and legs. The eLegs also uses electric motors to provide movement of the joints and monitors the user's arm motion to initiate a step.

Both devices use batteries to provide several hours of untethered operation on a single charge. While the ReWalk has been demonstrated climbing stairs, neither device has been demonstrated on rough or irregular terrain. Both devices target paraplegic users, and thus operating in a rigid position control mode is a viable option because the user cannot provide any motion of the legs. For paraparetics, a more compliant mode of operation is required, and it is not clear if either device can target this type of user. Both devices are undergoing clinical trials and neither device is currently available for personal use.

A device that has significant operational experience with able-bodied users is the hybrid assistive limb (HAL) [5]. The HAL is able to augment the user's capabilities by detecting intent with the use of EMG signals. A new version of this device, HAL-5 LB (Type C) targets paraplegic users [6]. This version of the HAL has only demonstrated a standing-up motion, not overground mobility. However, the HAL-5 LB (Type C) does have an actuator at the ankle, which the ReWalk and the eLegs do not. This actuator can make up for the lack of actuation in paraplegic users and assist in the balance task.

The Wearable Power-Assist Locomotor (WPAL) [7,8] is in a less mature phase of commercial development. This device has been demonstrated with paraplegic participants. The most significant difference between this device and the ReWalk and eLegs is that it has been demonstrated with only a walker as the balance aid, not crutches. The walker provides a significant support polygon for the user and produces more of a static tripedal gait.

[1]This material is based on research sponsored by the Office of Naval Research under agreement number N0014-09-1-0800 and N00014-10-1-0847. Any opinions, findings, and conclusions or recommendations expressed in this material are those of the authors and do not necessarily reflect the views of the Office of Naval Research.

Figure 1: IHMC Mina, a robotic orthosis for assisting mobility. Mina has four actuated degrees of freedom. To date it has enabled individuals with paraplegia to walk on flat ground up to 0.2 m/s.

The EXoskeleton for Patients and the Old by Sogang University (EXPOS) [9], was designed as a walking assist device for the elderly and for patients with muscle or nerve damage in the lower body. This device uses a wheeled caster walker to carry the actuators and computer system, and uses cables to transfer the actuator force to the exoskeleton joints. There is position feedback at the exoskeleton joints but no force sensing in the actuators. Force sensors on the leg braces are used to detect the user's intent. Because the caster walker component is integral to the device, operation and utility of this device is limited. A proposed exoskeleton by Zabaleta [10] will utilize compliant actuation for a robotic exoskeleton for rehabilitation, but details of the system have not been presented.

A number of robotic orthoses have been or are currently being developed for treadmill based operation. While these devices face some of the same challenges and share some of technologies as the overground devices, they are strictly limited to rehabilitation. It is with these devices that we are starting to see the use of compliant control and actuation. The PGO [11] LOPES [12] utilize force sensors on each actuator which can be used to provide torque control at the joint. One of the most utilized and studied treadmill based robotic orthosis device is the Lokomat [13,14]. While the first version was strictly a position controlled device, recent developments demonstrate that more compliant strategies are possible [15,16].

We have designed and built an overground robotic orthosis called Mina (Figure 1). This device is designed to provide mobility for paraplegic and paraparetic users. Because it utilizes compliant control actuators, it is able to change from rigid position control that is utilized with paraplegic users to assistive based control for paraparetic users. In its current state of development, Mina offers no advantages for paraplegic mobility over the ReWalk and eLegs, and is at a lower level of consumer readiness. However, Mina was developed to understand the device and training requirements for paraplegic mobility. This paper presents the evaluation of Mina with two paraplegic users.

In section II we provide details of the hardware design. We then discuss the method in which we generated the gait for operation with paraplegic users in section III. Next we discuss the evaluation of Mina in section IV. In section V and VI, we present and discuss the results and finally we present future work for this research in section VII.

II. DESIGN

Mina is a second generation [17] lower extremity robotic gait orthosis designed and built by researchers at the Institute for Human and Machine Cognition (IHMC). It has two actuated degrees of freedom per leg, hip flexion/extension and knee flexion/extension, for a total of four actuators. Mina does not provide any hip ab-/adduction or medial/lateral rotation of the leg. The torso section of Mina consists of a rigid back plate which has a curvature to match that of the human spine. Mina is designed to accommodate a range of body sizes. By using nested aluminum tubing as the structural links, adjustments are made to fit the user.

Mina attaches to the user at the torso and at three places on each leg: the thigh, shank, and foot. At the torso, there are two shoulder straps and a pelvis strap which secure the user's torso to the rigid back plate.

A tether provides Mina with power for the computer and motors, as well as Ethernet communication. While untethered operation is the eventual goal, having a fixed power source facilitates testing. In the following subsections the main components of Mina are introduced.

A. Actuators

Mina has four identical actuators (Figure 2). The actuators are capable of performing both position control and torque control. Each actuator consists of a DC brushless motor (Moog BN34-25EU-02) and a 160:1 harmonic drive (SHD-20 from HD Systems) gear reduction. The actuators are instrumented with two incremental encoders. The *motor encoder* measures the relative position between the motor shaft and the base of the actuator. This encoder (Avago HEDL-5640#A13) has a resolution of 2000 counts per revolution, which translates to $1.96 \cdot 10^{-5}$ rad/count at the output. The *output encoder* measures the relative position between the output of the actuator and the base of the actuator. This encoder is a Renishaw RGH-24 linear encoder. Its read-head is mounted onto the output of the actuator and the linear encoding strip is wrapped around a curved surface fixed to the base of the actuator. This encoder has a resolution of 1μm at a radius of 0.045m, which translates into a resolution of $2.22 \cdot 10^{-5}$ rad/count.

The motion of the output encoder matches the motion of the motor encoder except for any elastic deformation of the harmonic drive due to torque applied to the output shaft. The non-linear elastic properties of each actuator were characterized by applying known torques and measuring the resulting deflections using the difference between the two encoders. With a peak torque of about 60 Nm, the elastic deflection is about 0.0025 rad, for a resulting average stiffness

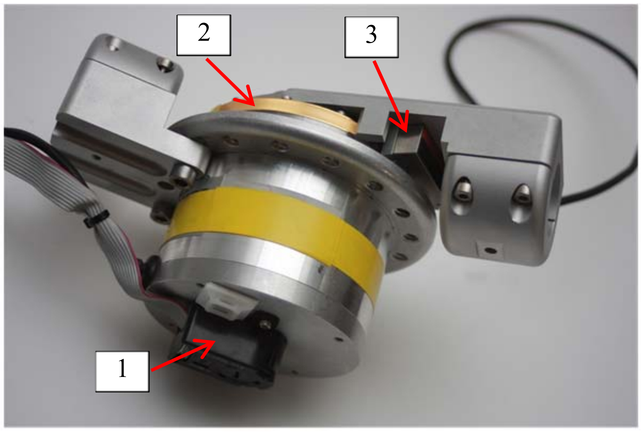

Figure 2: Actuator used in Mina. The block box (1) at the bottom of the picture is the encoder on the back shaft of the motor rotor. The gold strip (2) at the top of the picture, which is fixed to the base of the actuator, is the part of the output encoder. The read-head (3) of the output encoder is fixed to the output of the actuator. As the output moves, the read-head passes over the gold strip providing relative motion information

of about 24kNm/rad. During operation, the torque of the actuator is determined by calculating the deflection of the harmonic drive and then using a table look up to determine the corresponding torque.

Position control with high impedance is achieved by using only the motor shaft encoder. The deflection of the harmonic drive is considered to be negligible when the actuator is operating in position control mode. A simple controller using proportional plus derivative feedback from motor position to input motor current is used to track a desired position signal.

Torque control was achieved with a simple proportional plus derivative controller (Figure 3), where the error signal is the desired torque minus the applied torque and is used to determine the input current to the motor.

B. Computer and Electronics

Mina is controlled by an embedded computer system mounted onto the back plate. The embedded computer communicates with a desktop host computer via a tethered Ethernet cable. The embedded computer is running Solaris and all the control software is written in Real-Time Java. A stack of PC-104 boards is used to read the encoders, send PWM commands to the amplifiers, and handle analog and digital IO. The motor amplifiers used are Accelnet digital servo modules (ACM-180-20) by Copley Controls. Although they are capable of 20A peak current, they are configured for only 10A peak and continuous.

The embedded computer runs the control code and stores

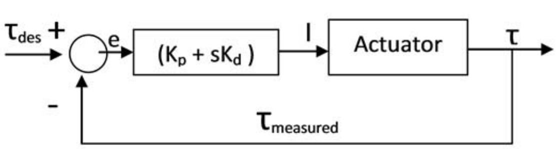

Figure 3: Diagram showing the feedback loop used to control the output torque of the actuator.

Figure 4: Placement of the C-shaped leg braces. The leg brace straps have been removed for clarity. Note that the thigh braces provide a rigid support at the back of the leg and the shank braces provide a rigid support at the front of the leg.

the trajectories that are used in paraplegic walking mode. The relevant state variables are sent from the embedded computer to a host computer in real time and displayed on a screen for the control operator, who is monitoring the user in Mina. Operational control signals are sent from the host computer to the embedded computer at a rate of 50Hz.

C. Braces

Each leg has two braces, one at the thigh and one at the shank (Figure 4). The braces exert the highest force on the leg during stance phase as Mina is supporting the gravitational load of the user. In this state, when the knee is even slightly bent, the contact force between Mina and user is at the tibia and the back of the thigh. To enable donning and doffing, the leg braces are designed with an opening of approximately 180 degrees, or an approximate C-shape.

To provide the most amount of comfort and support, the rigid parts of the braces are aligned to contact the leg at the point of the highest force. The thigh brace contacts the back of the thigh, and the shank brace contacts the tibia, or front of the shank. The leg is inserted into the opening of the brace and then secured with a Velcro strap.

D. Ankle and Foot

Mina was initially designed to provide ankle dorsiflexion/plantar flexion by a joint co-located with the user's joint (Figure 5). The joint contained a spring with a neutral position at five degrees of dorsiflexion. However, during testing with a paraplegic user, it was determined that this compliance was making it difficult for the user to balance with crutches. Therefore this compliant joint was replaced by a rigid joint, which eliminated motion of the foot plate relative to the

Figure 5: Original Mina ankle joint shown with spring. The joint is coincident with user's ankle joint and restricts rotation to dorsi/plantar flexion only. The neutral position of the spring is with five degrees of dorsiflexion. This flexible joint was replaced with a rigid joint that prevented movement of the ankle joint.

shank.

Moving distally down the leg the third point of contact between the user's leg and Mina is at the foot. The user's foot is inserted into an appropriately sized athletic shoe. The distal part of Mina frame terminates at a carbon fiber foot plate that is inserted into this shoe. The foot plate has some compliance so that during the end of stance phase the heel of the shoe can come off the ground while leaving the front part of the shoe in contact with the ground.

III. GAIT GENERATION AND OPERATION

Mina was used as a motion capture system to record trajectories from an able-bodied individual; these recorded motions are used in the paraplegic assistance mode. This method of gait generation was chosen because it allowed for a natural gait with a quick development cycle. In about 30 minutes, a new walk can be recorded and made ready for playback in paraplegic assistance mode.

A. Generating Walking Trajectories

Walking trajectories were generated by having an able-bodied person walk while wearing Mina, and recording the joint positions. During this process, the actuators were set to torque control mode. For the hip joints, the desired torque was set to zero so that Mina would follow the user's motions without affecting them.

For the knees, a slightly different torque value was used. When in torque control mode, there is generally a few degrees of offset between the user's joints and the device's joints because of compliance in the user's flesh and the braces of Mina. This means that when the user's knee joint is fully extended, the knee joint of Mina might not necessarily be fully extended. However, in paraplegic assistance mode, we wanted the knee joint to be fully extended during stance. In order to assist the knee joint to the fully extended position, the desired toque was set to be 10 Nm in extension. This torque ensured that the knee was fully extended during stance phase, yet small enough that the able-bodied user was able to overcome it during swing.

Toe off is a significant component of natural gait [18]. Because Mina does not have an ankle actuator, the walk used for the recorded gait had to be adjusted to account for this lack of actuation. The resulting walk was similar to walking on a slippery surface, where one has to minimize the ground reaction shear forces. The result is that the gait with an assist device, such as Mina, that does not have the same actuated degrees of freedom as a healthy person cannot have the same gait as a healthy person.

In natural walking, people minimize the ground clearance as part of a muscle energy conservation strategy. However, for a robotic orthosis, electrical energy conservation is not the same as muscle energy. Therefore, in order to guarantee that the toe does not stub on the ground prematurely, the user walked with exaggerated ground clearance in swing phase during gait recording. In human walking, there a is complex feedback loop between terrain sensing, joint position, and body position. Replicating this complex loop, especially the terrain sensing part will be studied in future work.

TABLE 1. WALKING SPEED FOR LARGE STEP

Leg Length (dist. from hip joint to ankle)	**Actual** Step Size	**Step Period**	**Walking Speed**
0.840m	0.24m	1.4s	0.18m/s
0.785m	0.28m	1.4s	0.20 m/s

After the recording phase, the trajectories were played back in paraplegic assistance mode with an able-bodied user trying to relax his lower limb muscles. From this playback, the best single gait cycle (stance and swing phase) was selected to use as a basis for the final walk. The joint angles at the end of this gait cycle were adjusted to match the starting joint angles, allowing the step to be played back in a smooth, endless loop. The joint angles were then copied to the other leg with a half cycle phase shift. This ensured that the left leg and the right leg execute the exact same step with the appropriate phase shift.

Three different walks were recorded, with step sizes ranging from zero (stepping in place) to what will be referred to as a *large* step. The precise value of the step size for a given walk depends on the leg length of the user. The quickest step period we tested Mina with was 1.4 seconds per step. The resulting walking speeds are presented in Table 1. Note that the recorded gait is sequence of desired joint angles (Figure 6). The resulting walking speed is a function how fast this sequence of desired joint angles is played and leg length of the user. The longer the user leg length the larger the actual step, and thus the faster the resulting walking speed. The fastest walk speed recorded was 0.2 m/s (see Table 1), which was limited by actuator performance rather than user capability.

B. Operation

The control operator uses a MIDI (Musical Instruments Digital Interface) interface (Behringer Model BCF2000) to control the walk when Mina is in paraplegic assistance mode. Some of the available control commands include turn actuators on, start walking, stop walking, and slow down. The operator has the ability to trigger a single step or continuous steps, as well as the ability to adjust the walking speed between 50% and 130% of the recorded speed. In addition, the operator can adjust the time the controller pauses between left and right steps. This is useful when the user is first learning how to walk in Mina.

During testing the control operator responds to verbal and

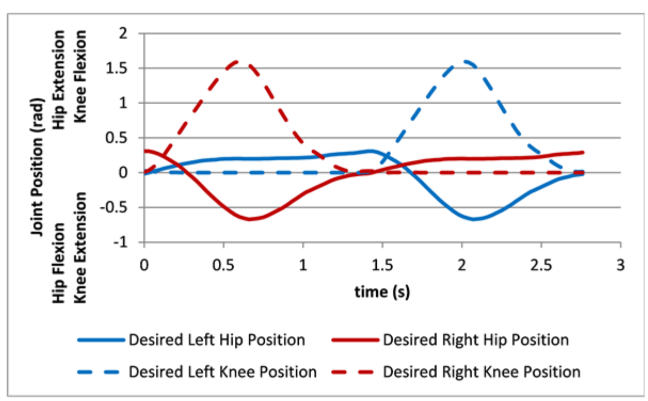

Figure 6: Desired joint angles for a right and then left large step.

gesture cues from the user to determine the appropriate command to send. The operator also acts in a safety capacity, and if the user is having balance issues or not positioning the crutches properly, the operator can stop the stepping.

For the purposes of this evaluation of Mina, having control initiated and maintained by the operator, and not the user, was sufficient and allowed the user to focus on learning how to walk and not worrying about any user interface issues. We acknowledge that this user interface is only a temporary implementation for the sole purpose of this preliminary evaluation. For a mobility assist device to be viable, giving the user full control is essential, and will be addressed in subsequent versions.

IV. Description of Evaluation

A study was conducted to evaluate Mina with paraplegic evaluators. The areas for evaluation include the donning and doffing procedure, the location of the braces and straps, the user interface, the joint trajectory for walking, the training procedure for users, required training time, the ease of use, and the ability for Mina to accommodate turning.

To eliminate the risk of falling, Mina was connected to an overhead rail system by a proof tested tether with very little slack. This safety system would only allow the user to “fall” a few inches before the safety tether became taut.

A. *Inclusion/Exclusion Criteria*

We required the evaluators to have an American Spinal Injury Association (ASIA) Impairment Scale [19] of either an A (Complete) or B (Incomplete). In order to ensure that the evaluator would be comfortable being in a standing position, we required that the evaluator have a Walking Index for Spinal Cord Injury (WISCI) level 9 (Ambulates with walker, with braces and no physical assistance, 10 m) or higher [20]. In future pilot studies we will relax the inclusion criteria to explore the range of potential users.

The testing protocol was approved by the IHMC Institutional Review Board. All evaluators were briefed on the risks related to the testing and were required to give informed consent. All testing sessions were monitored by a medical professional. Before each session, the medical professional would examine the evaluator for pre-existing bruises, open sores, skin lesion, or skin irritations that might interfere with the testing. After each session, the medical professional would again examine the evaluator to determine if using Mina caused any bruising, chafing, or skin irritations.

B. *Testing Plan*

Although the evaluators were able to walk prior to their injury, walking in Mina is much different than able-bodied walking. For complete paraplegics, there is no feedback of the ground reaction force and center of pressure in the feet. Additionally, SCI persons do not use their remaining proprioception feedback loop for balance as frequently as able-bodied persons because most of their time is spent seated. Finally, when walking in Mina, their arms become an integral part of balance and ambulation, walking like a quadruped rather than a biped. To make up for the lack of sensory feedback we placed a video monitor in front of the evaluator during initial training, which provided a real-time side view of themself.

For these reasons, training in Mina followed the steps listed below:

1. Stand in Mina between parallel bars.
2. Set step length to zero and step in place between parallel bars. This will help the user feel the load in the arms.
3. Set a small step length and walk between parallel bars. The evaluator should get used to being upright and moving arms in coordination with stepping.
4. Same conditions as above, but the evaluator is instructed to move hands in discrete motions, rather than slide hands along the bars.
5. Same conditions as above but the evaluator is instructed to only contact the parallel bars with the palm of the hand, limiting the shear force and eliminating the ability of pulling up on the ground. This is to prepare the evaluator for using crutches.
6. Same conditions as above but increase step length and stepping frequency.
7. Remove parallel bars, fit evaluator with forearm crutches. Set step length to zero and let the user get the feel of the shifting load.
8. Same conditions as above but increase step length and stepping frequency and remove video monitor feedback.

V. Result from Evaluation

Two evaluators tested Mina (see Table 2). Testing with evaluator 1 occurred in 9 sessions over the course of two months. Testing with evaluator 2 occurred in 9 sessions over the course of five days.

Table 3 summarizes the quantifiable testing results. The *time spent using the parallel bars* was defined as the time the evaluator spent standing and walking in Mina while using the parallel bars for support. The *time spent using crutches* was defined as the time that the evaluator spent standing or walking in Mina using crutches for support. The *total time in Mina* is

TABLE 2. EVALUATOR DESCRIPTION

Evaluator	Sex	Age	Years Since Injury	Level of Injury	ASIA Impairment Scale Classification
1	M	20	3	T10	A
2	F	26	4	T12	A

TABLE 3. EVALUATION TIME

Evaluator	Time spent using parallel bars (HH:MM)	Time spent using crutches (HH:MM)	Total time in Mina (HH:MM)
1	1:42	2:04	3:46
2	1:10	4:24	5:34

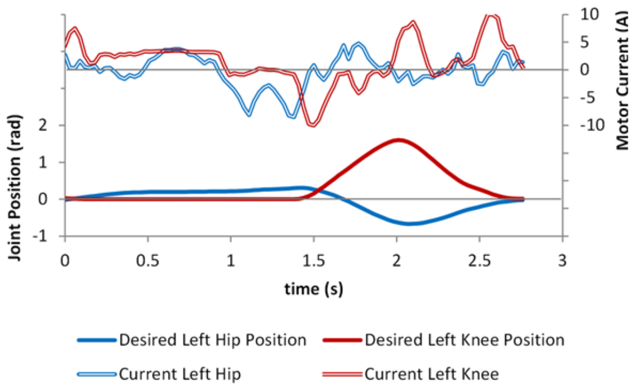

Figure 7: Plot showing motor currents and joint angles for the left leg of a gait sequence consisting of first right leg then a left leg step.

the sum of the previous two times.

The initial sessions with the first evaluator of Mina were spent fine tuning many aspects of the testing. We spent the first sessions determining the best procedure to don and doff Mina. We found that the braces and shoes dictated the procedure, and that redesign of both would facilitate the donning and doffing process.

There was no formal criterion for transitioning from parallel bars to the crutches. It was not until after the first evaluator switched to crutches that we concluded that practice with the parallel bars does not entirely carryover to crutches. Our rough criterion for transition was that the evaluator was able to walk in the parallel bars without sliding his or her hands on or pulling on the parallel bars. While a study could be conducted to determine the effectiveness of parallel bar training in preparation for forearm crutch use, we feel that optimizing the short amount of time spent in parallel bar use would have limited overall effects in the outcome of mobility.

One of the most significant changes made to the procedure was adjusting the step sizes. We initially started with a step length of around 60 cm. However, during testing with this very large step, the evaluator's center of mass was between the two feet during double support. Because the trailing leg was still loaded as it initiated the next swing phase, the evaluator started to fall backwards. In response to falling, the evaluator would pull on the parallel bars to move his center of mass forward, over the stance foot. This problem could be alleviated by one of three remedies: 1) a dynamic gait, 2) an ankle actuator for toe off, or 3) smaller steps. The last option was selected and the first two options will be explored in future research. After the third session with the first participant, we reduced the step size to a small (approximately 11 cm) step, and a large (approximately 26 cm) step.

Figure 7 shows the motor current required to track the desired joint trajectory during operation with one of the evaluators. Because the tracking error was less than +/-0.04 rad at the hip and less than +/-0.06 rad at the knee, only the desired joint positions are shown. This figure confirms that the motors are rarely at peak current (10A).

Progressing to using forearm crutches as the only balance aid proved to be more of a challenge with the evaluators than with abled-bodied users simulating paralysis.

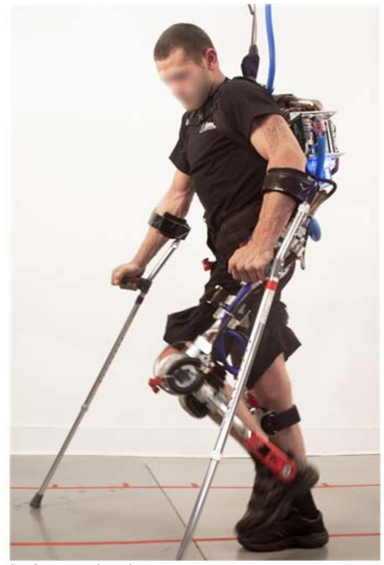

Figure 8: Mina during evaluation

Because we had debugged the training and coaching process with the first evaluator, the second evaluator progressed more rapidly through the testing plan.

By the end of testing, both evaluators were able to easily walk with forearm crutches (Figure 8). We coached the evaluators to think of themselves as a quadruped. For quadrupeds, a common slow walking gait is a regular crawl, consisting of a hind, ipsilateral front, contralateral hind, and contralateral front sequence. We instructed the evaluators to walk using the same sequence, where the crutches were their front legs, and their legs were the hind legs. Although we did not require the evaluators to use this gait, both reported that this sequencing felt comfortable and natural.

We qualitatively evaluated the cognitive effort required to use Mina by having a conversation with the evaluator while walking with Mina. Both evaluators eventually reached the point where they could maintain eye contact and have a conversation while walking in Mina.

We tested the static standing balance stability by having evaluator 2 catch and throw a ball while standing on both legs and using one crutch for balance. We varied the throw so that she had to move her torso and arm to be able to reach the ball to catch it. She was able to perform this task without losing balance. Standing balance stability is required for activities of daily living such as pressing elevator buttons, opening a door, or having a drink in a standing-room-only bar. Future evaluations would include these types of functional tests as well as balance stability during walking.

Both evaluators were able to turn in place and while walking in Mina. To turn while walking, the user would twist his or her torso during single support, using the two crutch supports to provide the counter torque. To turn in place, the user would press down with the crutches and hop and turn with both feet.

For both evaluators, one of the most difficult parts of learning to use Mina was determining the body, angle in the sagittal plane, relative to gravity during single support. Mina rigidly positions the torso relative to the legs, and the entire body can be considered an inverted pendulum. Although the ankle joint is rigid, there is still compliance in the foot plate and shoe, allowing the user to move his or her center of mass forward and backward.

The user must learn how to position his or her body at the point of heel strike. If the user is leaning too far backward, then the upcoming swing leg will still be loaded at the time of swing, causing the user to fall backward. If the user is leaning too far forward, the foot will contact the ground before the swing motion is complete, resulting in significantly reduced step size.

We coached the evaluators to lean just enough so that the heel of the upcoming swing leg was slightly off the ground at the end of the double support phase. Because the evaluator has no sensory feedback to know if his or her foot is off the ground, during training the evaluator used the side view camera angle to see when his or her heel was off the ground.

Neither evaluator found walking in Mina to require significant physical exertion. During testing, the medical professional would monitor physical exertion by measuring pulse, respiration rate, skin color, and perspiration level. Once the evaluator was experienced in walking in Mina, there was hardly any perceivable change in physiological response between standing and walking.

No bruising or chaffing was observed on either evaluator. One evaluator had slight swelling around the knee after standing in Mina for over an hour. Increased blood flow in the legs was observed in both evaluators.

VI. Discussion

We presented the robotic orthosis Mina to assist a paralyzed person in walking. We evaluated Mina with two paraplegic evaluators and demonstrated that Mina is currently capable of providing mobility for paraplegic users on flat ground at slow walking speed. While Mina is not yet capable of being used to enhance a person's quality of life in real world settings, we are confident that such devices are technically feasible. Our goal is to develop a device that enables paraplegic individuals to ambulate on unimproved roads and trails, through typical urban environments, and just about any place on the Earth that an able-bodied individual can go.

We determined that true evaluation of Mina for paraplegic mobility assistance could only be done with paraplegic users, not able-bodied users simulating paralysis. Even though Mina currently operates in a high impedance trajectory-tracking mode, it is difficult for an able-bodied user to stay completely relaxed in the device and instead finds their muscles actively trying to walk and balance. Also, an able-bodied person receives sensory feedback, such as ground reaction force feedback, which a paraplegic user does not receive. Therefore, we believe that true evaluation of such devices with the aim of restoring gait in paraplegic individuals can only be truly evaluated with such individuals.

In evaluating Mina, we observed that all users required some amount of training and practice, and that more training and practice was required for paraplegic users than able-bodied users. As with any new activity that requires coordinated motion, it takes practice to become proficient. Walking with forearm crutches in Mina is akin to cross-country skiing from a coordination point of view. For users with no proprioception in their lower extremities, the task of coordinating the arm motion with the leg motion requires more and different practice than for able-bodied users. One difference in training for paraplegic users was providing them with video feedback of what their legs are doing. We hypothesize that providing these users with some form of tactile feedback that accounts for some of the missing proprioception will not only decrease the learning time, but increase performance and stability.

VII. Future Work

To date, Mina has been able to assist two paralyzed users while walking at relatively low speed on flat ground in a laboratory setting. Next steps include increasing the speed of walking, walking over rough terrain, on hiking trails, and in urban environments with stairs and narrow passageways.

One of our first areas of development will be providing users with real-time foot ground reaction force information. We will then test if this accelerates the learning period and increases their stability when using Mina.

Mina currently plays back pre-recorded joint trajectories and the user maintains balance with his or her arms and upper body. We are currently developing various intuitive interfaces to give the user operational control of the device. There are many options for such operational control. For example, a step could be initiated from motion or force sensed in the feet and/or forearm crutches, or from gestures provided by upper body motion. Step height and where to step when walking on rough ground could be guided by similar gestures or perhaps even by eye tracking.

To date Mina has used high gain rigid joint trajectory tracking during playback. In our bipedal walking robots we advocate compliant control that avoids rigid joint trajectory tracking. Compliant control allows for robustness to unmodeled terrain variations. We plan to investigate similar compliant control techniques with Mina. We believe that with improvements to user feedback, intuitive user interfaces, and compliant control techniques, users should be able to comfortably walk over relatively rough terrain.

Having two evaluators with different body size and multiple testing sessions highlighted some of the areas for improvement of the mechanical design of Mina. The first area is the donning and doffing process. This process required the assistance of a couple of people. With design changes to the braces, we are confident we can produce a design that requires no assistance for users capable of self-transfer from a wheelchair.

While a device like Mina can transform the mobility options for people with paralysis, there are other potential benefits to the users that need to be explored. There is the potential to use Mina as part of a gait rehabilitation intervention for spinal cord injured patients. Trials for this would be to compare the functional mobility restoration for conventional therapy to that therapy using Mina. Another area for study is the health benefits associated with the use of Mina to patients confined to a wheelchair. This study would examine areas including pressure sores, bone density, circulation, digestion, and could even extend to mental health.

What leads us to believe that this type of device and mobility is desired by the target population is what one of the evaluators said during the testing:

> "What I want is to be able to reintegrate back into the walking society. Because we are not rolling creatures. The exoskeleton [Mina] will bring me back to what humans are…we're walking creatures."

Acknowledgment

We would like to thank our two evaluators for their invaluable feedback in testing Mina. We would also like to thank the team of medical professionals that volunteered their time to monitor our evaluation sessions.

References

[1] M. Berkowitz, P. O'Leary, D. Kruse, and C. Harvey, *Spinal cord injury: An analysis of medical and social costs*. New York, New York: Demos Medical Publishing, 1998.

[2] S. Goemaere, L. M. Van, N. P. De, and J. M. Kaufman, "Bone mineral status in paraplegic patients who do or do not perform standing," *Osteoporosis International : a Journal Established As Result of Cooperation between the European Foundation for Osteoporosis and the National Osteoporosis Foundation of the USA*, pp. 138-43, 1994.

[3] T. Sumiya, K. Kawamura, A. Tokuhiro, H. Takechi, and H. Ogata, "A survey of wheelchair use by paraplegic individuals in Japan. Part 2: Prevalence of pressure sores," *Spinal Cord*, pp. 595-598, 1997.

[4] R.L. Ruff, S.S. Ruff, and X Wang, "Persistent benefits of rehabilitation on pain and life quality for nonambulatory patients with spinal epidural metastasis," *Journal of Rehabilitation Research and Development*, pp. 271-280, 2007.

[5] T. Hayashi, H. Kawamoto, and Y. Sankai, "Control method of robot suit HAL working as operator's muscle using biological and dynamical information," *IEEE/RSJ International Conference on Intelligent Robots and Systems*, pp. 3063-3068, 2005.

[6] A. Tsukahara, Y. Hasegawa, and Y Sankai, "Standing-up motion support for paraplegic patient with Robot Suit HAL," in *IEEE Conference on Rehabilitation Robotics*, 2009, pp. 211-217.

[7] T. Kagawa and Y. Uno, "A human interface for stride control on a wearable robot," in *IEEE/RSJ International Conference on Intelligent Robots and Systems*, St. Louis, MO, 2009, pp. 4067-4072.

[8] T. Kawaga and Y. Uno, "Gait pattern generation for a power-assist device of paraplegic gait," in *18th IEEE Int. Symposium on Robot and Human Interactive Communication*, Toyama, Japan, 2009, pp. 633-638.

[9] K. Kong and D. Jeon, "Design and Control of an Exoskeleton for the Elderly and Patients," *IEEE/ASME Transactions on Mechatronics*, p. 428, 2006.

[10] H. Zabaleta et al., "Exoskeleton design for functional rehabilitation in patients with neurological disorders and stroke," in *International Conference on Rehabilitation Robotics*, Noordwijk, Netherlands, 2007, pp. 112-118.

[11] Z. Feng et al., "Biomechanical design of the powered gait orthosis," in *IEEE International Conference on Robotics and Biomimetics*, Sanya, China, 2007, pp. 1698-1702.

[12] R. Ekkelenkamp, J. Veneman, and H. van der Kooij, "LOPES : Selective control of gait functions during the gait," in *IEEE 9th International Conference on Rehabilitation Robotics*, Chicago, 2005.

[13] G. Colombo, M. Joerg, R. Schreier, and V. Dietz, "Treadmill Training of Paraplegic Patients using a robotic Orthosis," *Journal of Rehabilitation Research and Development*, pp. 693-700, 2000.

[14] G. Colombo, J. Matthias, and V. Dietz, "Driven gait orthosis to do locomotor training of paraplegic patients," in *Engineering in Medicine and Biology Society, 2000. Proceedings of the 22nd Annual International Conference of the IEEE* , Chicago, IL, 2000.

[15] R. Riener et al., "Patient-cooperative strategies for robot-aided treadmill training: first experimental results," *IEEE Transaction of Neural Systems and Rehabilitation Engineering*, pp. 380-394, 2005.

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

[17] H.K. Kwa et al., "Development of the IHMC Mobility Assist Exoskeleton," in *Proceedings of the 2009 IEEE International Conference on Robotics and Automation*, Kobe, Japan, 2009, pp. 2556-2562.

[18] David A. Winter, *Biomechanics and motor control of human movement*, 2nd ed. New York, New York: Wiley, 1990.

[19] F. M., Bracken, M. B., Creasey, G., Ditunno, J. F. Maynard, W. H. Donovan, T. B. Ducker, S. L. Garber, and C. H. Tator, "International Standards for Neurological and Functional Classification of Spinal Cord Injury," *Spinal Cord*, pp. 266-274, 1997.

[20] B. Morganti, G. Scivoletto, P. Ditunno, J. F. Ditunno, and M. Molinari, "Walking index for spinal cord injury (WISCI): criterion validation," *Spinal Cord: The Official Journal of the International Medical Society of Paraplegia*, pp. 27-33, 2005.