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Development of the IHMC Mobility Assist Exoskeleton

Hian Kai Kwa, Jerryll H. Noorden, Mathew Missel, Travis Craig,
Jerry E. Pratt, Peter D. Neuhaus
Florida Institute for Human and Machine Cognition
Pensacola Fl 32502, USA

***Abstract*— The IHMC Mobility Assist Exoskeleton is a robotic suit that a user can wear for strength augmentation or gait generation. This first generation exoskeleton prototype focuses on providing walking assistance to persons with lower extremity paralysis. The main goal is to successfully enable a person that cannot walk without assistance to walk in a straight line a distance of 15 feet. When in disable assist mode this prototype will rely on the user to provide balance control, and thus an external means for balancing will be required, such as crutches or a walker. Power and control is off board and supplied to the Exoskeleton by means of a tether. Rotary Series Elastic Actuators (RSEAs), which have high force fidelity and low impedance were designed to power the joints. This paper describes the design, test results, future work and potential applications of the exoskeleton.**

I. INTRODUCTION

ALTHOUGH many exoskeletons have been developed, most are designed for performance (strength and endurance) augmentation, Performance Augmentation Exoskeletons. These exoskeletons are typically pseudo-anthropomorphic in design, where the exoskeleton and the user's joints do not coincide exactly and the only connection point is at the end of the limb. This makes them unsuitable for applications where the exoskeleton must move the user's limbs for them; for example with Disabled Assist Exoskeletons. The IHMC Mobility Assist Exoskeleton (see Figure 1) is designed for the latter application and can move with a passive lower extremity pilot, as would be the case if the user were paraplegic.

A Disabled Assist Exoskeleton introduces a new set of challenges and considerations compared to Performance Augmentation Exoskeletons. User intent cannot be detected using force sensors attached to the legs, which is a common method for controlling exoskeletons. Complete paraplegics suffer from an impairment in motor and sensory function of the lower extremities and therefore must obtain feedback on position and loading of the legs through external means. They also face further challenges, including bone and muscle loss as a result of lack of use and loading of the lower extremities.

The IHMC Mobility Assist Exoskeleton aims to address these issues by providing a novel control and feedback system to the user, allowing the user to walk with assistance from the exoskeleton. While walking in the exoskeleton the user's legs will be loaded with his/her body weight and the rate of muscle and bone degradation will be reduced or possibly reversed.

For this first phase of the IHMC Mobility Assist Exoskeleton project, the goal is to provide assistance such that a person with a lower extremity paralysis can walk 15 feet in a straight line.

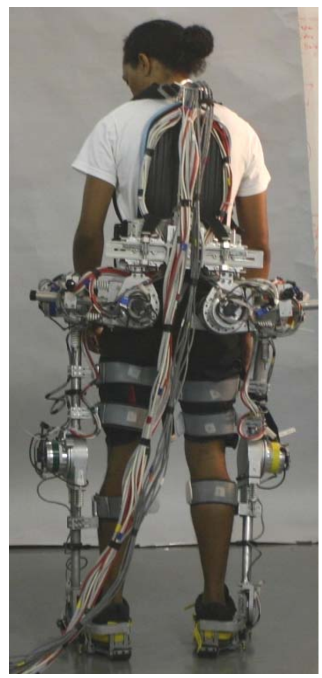

Figure 1: IHMC Mobility Assist Exoskeleton with user. The exoskeleton has three actuated degrees of freedom per leg: hip adduction, hip flexion, and knee flexion. There are also two passive degrees of freedom per leg: hip internal rotation and ankle flexion. The bundle of wires going to the exoskeleton carries motor power and sensor data.

II. BACKGROUND

There has been a significant increase in exoskeleton research, design and development in the last decade. Several organizations all over the globe have designed and built impressive exoskeletons for specific applications and goals, differing significantly in performance and technology used. They can be broadly categorized into two groups: exoskeletons for assistance and rehabilitation, and exoskeletons for strength and endurance augmentation.

The Exoskeleton for Patients and the Old by Sogang University (EXPOS)[1] was designed as a walking assist device for old people and for patients with muscle or nerve damage in the lower body. This device uses a wheeled caster to carry the actuators and computer system, and uses cables to transfer the actuator force to the exoskeleton joint. 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 intent.

The quasi-passive exoskeleton developed by Walsh demonstrated the need to match the user's joints and degrees of freedom in order allow for a natural gait [2]. It also shows that a fully force controllable lower extremity prototype can be used to explore different active as well as passive based controllers.

The "Soft" exoskeleton design by Caldwell demonstrates the potential for lightweight compliant actuation [3]. However, using a compressible fluid in the actuators lowers the position control bandwidth and reduced the potential for full disable assistance.

The Hybrid Assistive Limb (HAL)[4] exoskeleton was developed at University of Tsukuba, Japan as a full body exoskeleton for both walking assistance and strength augmentation. It is capable of driving the user's limbs using a position lookup table, and it also incorporates myoelectric sensors to monitor the muscle activity of the user. The information on muscle activity is then used to predict the intent of the user.

ReWalk is a lower extremity robotic suit developed by Argo Medical Technologies in Haifa, Israel [5]. It uses battery powered DC motors to actuate extension in the hip and knee and allows a paraplegic user to walk using crutches to help keep balance. It also incorporates a rotational sensor in the chest to detect the torso angle and adjust the legs accordingly. The ReWalk is strictly a disable assist device and cannot be used in a performance enhancement role.

Hocoma in Switzerland developed the Locomat, an anthropomorphic lower extremity exoskeleton used for rehabilitation [6]. It is suspended by a frame over a treadmill and actively moves the patient's legs through a walking motion over the treadmill while supporting some of their weight. The Locomat is a gait retraining device that is limited to rehabilitation use only through simulated over ground walking on a treadmill.

The Berkeley Lower Extremity Exoskeleton (BLEEX)[7] was developed at University Of California, Berkeley and has spawned several later generation exoskeletons like the Berkeley Bionics ExoHiker and the ExoClimber. These exoskeletons are primarily for augmenting human endurance while walking and carrying heavy loads.

Sarcos, in Salt Lake City, Utah, developed its full body exoskeleton to augment human strength and carrying capacity [8]. It uses hydraulic actuators and load cells for force feedback.

Both BLEEX and SARCOS are pseudo anthropomorphic in design, and the joints of the exoskeleton do not necessarily follow exactly the joints of the user. They were designed to have a load path connecting ground to the load that the user is carrying; for SARCOS the load is carried by the user's hands and for BLEEX it is mounted like a backpack. They are not designed to move the user's limbs for them.

The RoboKnee was developed by Yobotics in Boston, Massachusetts [9]. It is essentially an exoskeleton for the knee joint and uses a series elastic actuator to control the torque at the joint. Desired knee torque was computed based on force measurements between the foot and the ground. This exoskeleton was developed to test the feasibility of force control using series elastic actuators on an exoskeleton.

The latter three examples are exoskeletons developed for strength or endurance augmentation, while the earlier ones are for gait assistance and rehabilitation. As these examples show, there is a very broad range of objectives, approaches and technology used in modern exoskeleton projects.

III. Rotary Series Elastic Actuators

In order for an exoskeleton to be able act in an enhancement role or assist role, the actuators must be capable of either torque control, position control, or a combination of the two. Series Elastic Actuators [10] are an ideal choice for actuation in exoskeletons, because they offer high fidelity impedance control.

In a Series Elastic Actuator a compliant element is placed in series with the actuator output, and the force is calculated from measuring its compression. The advantage of this system is that it gives very accurate force feedback and has a low impedance. The disadvantage is that is has a relatively low bandwidth at high forces. This is caused by the elastic element taking time to compress when a force is applied, thus adding a delay to the force feedback signal.

Series elastic actuators can very closely approximate an ideal force source [11]. This is advantageous in our application because it allows the control system to apply a force at a joint independent of its position and velocity.

Clinical Gait Analysis data from [12] was referenced to determine the motor torque requirements for each powered joint on the IHMC Mobility Assist Exoskeleton. From the human joint gait analysis data the peak power and torque requirements as well as the RMS power could be determined. For the hip and knee, the peak torque during the stance phase was 40 Nm. The torque also peaks in the opposite direction at heel strike, but at this point the power is dissipative.

From these specifications we designed a Rotary Series Elastic Actuator (RSEA) using a Moog BN34-25EU-02 brushless DC motor paired with a 1:100 gear ratio harmonic drive from HD Systems (SHD-20). For the compliant element we designed a mechanism that uses a steel cable to transform the rotational motion of the gearbox output into linear motion. The force is then applied to a linear die spring which then exerts a moment on the output. Linear die springs were used because of their very predictable force-to-displacement function and good energy storage to weight ratio.

The RSEA developed (see Figure 2) for the exoskeleton has a torque limit of 80 Nm, a velocity limit of 6.8 rad/s, and a torque and position resolution of 0.02 Nm and 0.002 degrees, respectively (Theoretically calculated based on hardware specifications). It is ideally suited for this application because of its low impedance and high force and position fidelity. Its main drawback is a low torque control bandwidth of approximately 10 Hz at amplitudes higher than 15Nm. At lower torques, the a RSEA can track up to 30 Hz. The main reason for the low bandwidth is that at large torque amplitudes we start to see saturation of the motor velocity and amplifier peak current. Because human walking does not require high frequency torque following at large torque outputs, the low bandwidth was a problem.

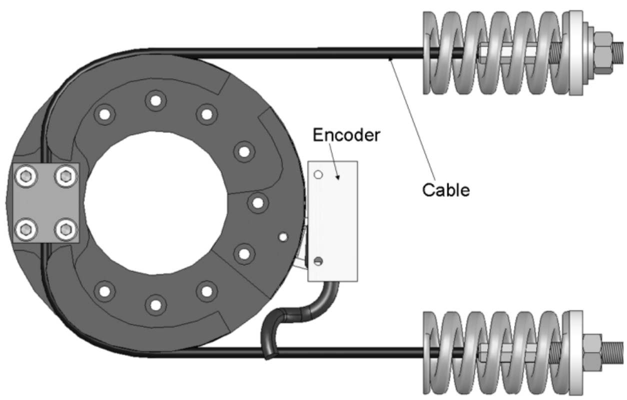

Figure 2: Schematic of the RSEA elastic mechanism design, shown in the zero force position. The encoder reads a scale tape on the curved cylindrical surface. Note that the actuator housing and bearing have been omitted for clarity.

IV. Exoskeleton Design

The primary goal of the IHMC Mobility Assist Exoskeleton is to enable a person, who was previously unable to walk, to walk in a straight line for a minimum of 15 feet.

Commonly, there are two approaches to exoskeleton design: the first is pseudo anthropomorphic; the second is anthropomorphic, where the joints are co-located as close as possible and where there are connection points at the thigh and shank as well as the foot. Because the exoskeleton is used to control the user's limbs in a disable assist mode, the former approach is not suitable for disable assist. For the latter approach, a wide range of adjustability for various body dimensions is needed to match the exoskeleton's joints as close as possible to that of the user's.

In this initial prototype we do not address balancing, and there are no means for detecting the orientation of the exoskeleton and the user. The user is required to provide balance control using torso movements and a pair of arm crutches. Furthermore, not all DOFs are actuated, and assisted walking is attempted using the absolute minimum requirements to accomplish this.

TABLE I
LIST OF THE EXOSKELETON DEGREES OF FREEDOM AND THEIR CONTROL METHOD.

Joint	Control Method	Range of Motion
Hip Pitch (extension)	Actuated	+42º (fwd), -30º(back)
Hip Roll (adduction)	Actuated	+25º (out), -30º (in)
Hip Yaw (rotation)	Passive	± 10º
Knee Pitch (extension)	Actuated	+0 º, -90 º
Ankle Pitch (dorsiflexion / plantarflexion)	Passive	+20º (up), -35º (down)
Ankle Roll (inversion / eversion)	Constrained	N/A
Ankle Yaw (rotation)	Constrained	N/A

Co-locating the actuator axes to the user's joint axes is difficult due to the dynamic behavior of human joints. In our exoskeleton we use fixed, single axis joints such that the actuator axis passes through the approximate center of the user's corresponding joint.

The IHMC Mobility Assist Exoskeleton (see Figure 3) has a total of ten Degrees Of Freedom (DOF), five per leg, six of which are actuated and four of which are passive (see Table I). The joints on each leg are connected in series, starting at the hip and going down to the ankle.

The first three DOFs form the hip joint; these are the hip yaw (lateral/medial rotation), hip roll (ad/abduction), and hip pitch (flexion/extension). The exoskeleton can be adjusted for the user's body size so that the axes of these three DOFs intersect at the user's actual hip joint. The hip yaw DOF, a passive joint, is achieved using a curved roller bearing which locates the center of rotation approximately at the user's hip joint. The curved roller bearing allows the entire exoskeleton leg to rotate about a vertical axis through the approximate center of the user's hip joint. This DOF is spring loaded to the center point, with the leg facing forwards. Due to the small range of motion of the spring loaded hip rotation (± 10º), moderate misalignment is acceptable.

The next hip DOF is the hip roll, which is actuated with the RSEA designed for this project. This DOF provides the user with the ability to side step and helps in turning. The final hip DOF is hip pitch. This DOF is also actuated with the RSEA.

The next DOF is the knee flexion/extension, which is actuated with the RSEA. The knee joint is connected to the hip pitch joint with telescoping round tubes. This allows translation and rotation adjustment of the knee joint relative to the hip joint.

The final DOF in the linear chain is the ankle joint. This DOF is unidirectionally spring loaded to provide the necessary dorsal flexion needed for ground-toe clearance during walking.

To prevent overextending the user's joints, the exoskeleton's joints have been fitted with mechanical limit stops tested to withstand peak torque capabilities of the motor and amplifier setup.

To ensure comfort and to align the actuator joints to those of the user, numerous position adjustments for the exoskeleton joints and the body connector braces were incorporated. The ranges of adjustments were chosen to fit the 10^{th} to 90^{th} percentile of users [13] [14].

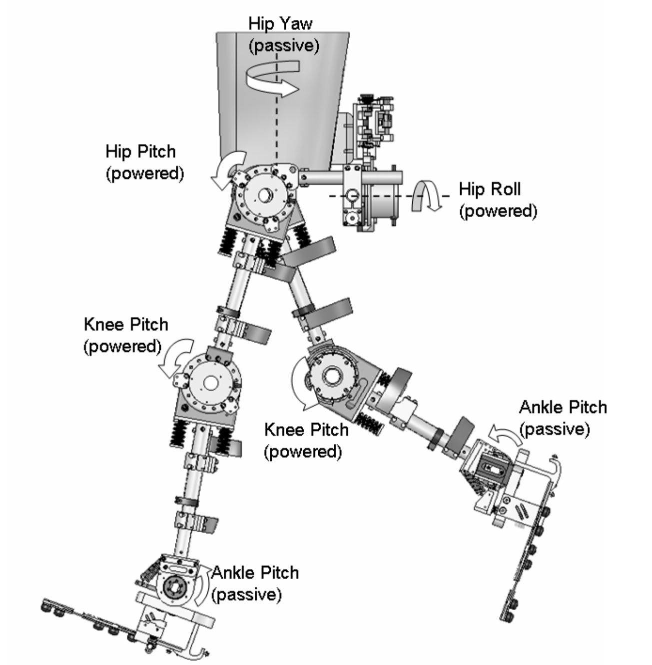

Figure 3: Side view of the IHMC Mobility Assist Exoskeleton showing the powered and passive degrees of freedom.

Other than position and force sensors on the actuators, the exoskeleton also has foot switches to detect if the foot is on the ground. There are two switches per foot: one at the heel and one at the toes. This enables the control system to determine whether the exoskeleton is in the single support (only one foot is loaded) or double support (both feet loaded) and approximately how the load is distribute on the foot.

The exoskeleton is able to support its own weight by providing a load path through the leg and feet structures to the ground. The Exoskeleton does not support the weight of the user; the user's weight is transmitted through his/her bone structure to the ground. The objective is to subject the user's bone structure to the normal gravitational induced loads experienced during walking, thereby reducing bone loss caused by disuse.

V. Electronics Setup

The exoskeleton is controlled by an off-board control system, and powered by means of tether. There are two wire bundles (signal cables and motor power cables) that run between the exoskeleton and the computer and are kept separate to minimize noise interference and signal corruption between these two bundles. An embedded PC-104 with custom circuit boards to connect to the amplifiers is used as the exoskeleton's computer system. The motor amplifiers used are Accelnet digital servo modules ACM-180-20 by Copley Controls. These amplifiers limit the output torque of the actuators; the peak current output from the amplifiers is 20A for a duration of 1 sec, and the maximum continuous current is 10A. The motor is rated for 37A peak and 10.5A continuous. With a 100:1 gearbox this results in an output torque of 80 Nm peak and 40 Nm continuous and a velocity limit of 6.8 rad/s.

Incremental optical rotary encoders were used to read the joint positions. The encoders (Avago HEDL-5640#A13) are mounted on the back of the electric motors and read the position before the gear reduction. The optical encoders used, have a resolution of 2000 counts / rev; ($2x10^5$ counts / rev at geared shaft output). Actuator torque is measured using a Renishaw RGH-24 linear encoder with a reference mark for zeroing (initializing) the force reading.

VI. Control Schemes

Rotary Series Elastic Actuators are capable of both low and high impedance within the rated bandwidth and torque output of the actuator. This allows for a broad range of control schemes, including torque control, position control, impedance control or combinations thereof. As a performance enhancement device, the exoskeleton is operated in torque control mode, and as a disable assist exoskeleton position control mode is used.

Torque/Force Control.

In torque control mode the actuator attempts to apply a torque equal to a desired torque, regardless of output position or velocity. A measure of fidelity of torque control for the actuator can be determined when the actuator is commanded to produce zero torque. Here, any externally applied torque on the actuator generates a negative force feedback, to which the control system responds with torque in opposite direction. The motor then rotates out of path of resistance (out of the way) as a result; making the actuator behave like a passive bearing joint. With the combination of spring and position encoder, the actuators can detect torque in increments of 0.012 Nm and can exert a peak torque in excess of 75 Nm.

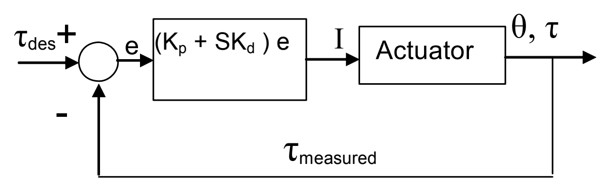

Figure 4: Torque control loop in the actuator control system.

The block diagram for the torque control loop is shown in Figure 4. The measured torque, $\tau_{measured}$, is a function of the spring deflection of the Rotary Series Elastic Actuator. The current, *I*, fed into the motor is calculated from a proportional plus derivative controller based on the torque error.

Position Control

In position control mode the exoskeleton positions the user's legs, moving them through a pre-programmed trajectory. In this mode, a feedback controller is used to have the position of the joint track a desired position. A block diagram for the feedback controller is shown in Figure 5:

For position control, a PD controller is used to generate a reference torque, which is then used as the input to the torque feedback controller. An inner torque control loop is used in the system to ensure that the torque commanded by the position controller is accurately tracked. Since torque is measured at the output of the gearbox this model does not need to account for motor and gearbox friction. The actuator is essentially used as a torque source/generator that is controlled by a position controller. Because of this nested controller structure, the position controller performance has a lower bandwidth than the torque controller.

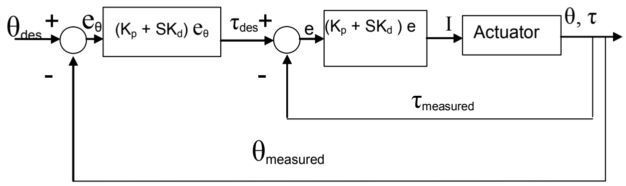

Figure 5: Control system diagram for an actuator. The torque feedback is fed to an inner control loop while the position is fed back to the overall position controller.

Using a 4.5 kg weight at the end of a 40cm moment arm we simulated the inertia of a human calf and foot to test the position control capability of the actuator. Under these conditions the bandwidth of the position controller is 1.9 Hz. Our final control system is slightly under damped resulting in some overshoot and oscillations in the test but when implemented in the exoskeleton the user's leg joint will add damping to the system.

Position and Force Control Combinations

Because all actuators on the exoskeleton have their own dedicated amplifier and control parameters, they can all be controlled independently. This allows for a broader range of applications. For patients with localized lower extremity injury, or in cases where only one leg is problematic, or even when both lower extremities are not subjected to the same conditions (paralyzed in only one leg), the exoskeleton's functionality can be adapted to conform to the user's specific condition(s). Torque control and position control modes can be dynamically triggered during the walking gait by numerous trigger signals such as a hand held button, or even a combination of the 4 foot switches depending on the specific phase of the walking gait.

VII. Potential Applications

Because of the high fidelity adjustable impedance of the Rotary Series Elastic Actuators, the IHMC Mobility Assist Exoskeleton can be used in a range of research applications.

Zero Assistance Control

A successful exoskeleton will assist the user when able and otherwise stay out of the way and not impede the user. To verify this metric, the IHMC Mobility Assist Exoskeleton was programmed to perform gravity compensation only to relieve the user of the weight of the hardware, but not to provide any assistance otherwise. The user then walked with a regular gait while the user's joint angles were recorded using the ShapeWrap II motion tracking system from Measurand, Inc. (a motion tracking suit that records motion data).

With the IHMC Mobility Assist Exoskeleton in this mode the user can wear it and walk around normally while feeling little or no resistance. Figure 6: shows a comparison of the user's hip and knee flexion joint angles while walking with and without the IHMC Mobility Assist Exoskeleton in zero assistance control mode.

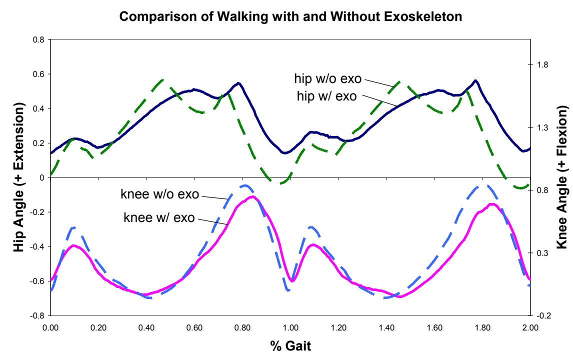

Figure 6: Comparison of hip and knee joint data without exoskeleton and with exoskeleton in zero assistance mode for regular walking. Joint data was taken with ShapeWrap II motion tracking system from Measurand, Inc. Scale and offset of joint angles is repeatable, but not accurate. Data shows that wearing the exoskeleton in zero assistance mode does not significantly affect gait.

Performance Enhancement

In performance enhancement mode, the control objective is to counter the gravitational forces subjected to the user, as well as any load the user may be carrying. In this application, the actuators are set to torque control mode. The desired torque is a function of the gait phase of the leg (swing or stance), the configuration of the joints, and the desired amount of assistance.

Gait Rehabilitation

The IHMC Mobility Assist Exoskeleton has great potential for gait rehabilitation because of the ability for high fidelity impedance control incorporated into a mobile suit, allowing for over ground gait training. An example application is rehabilitation for hemiparetic stroke survivors. It is possible to program the behavior of each joint separately in software, thus different combinations of position and force control can be utilized. For this case, one leg would be under forced gait control and the other leg would be under zero assistance control. Another potential application is for rehabilitation purposes where a patient regains partial muscle control; a position controller could be used with smaller gains such that the user's muscles have to apply a portion of the forces needed for walking. This resistance can then be adjusted as the patients builds up muscle integrity in their leg(s).

Disable Assist

The IHMC Mobility Assist Exoskeleton is primarily designed to be able to assist a user with lower extremity paralysis. The exoskeleton generates the position trajectories for the joints while the user provides balance through motions of the torso and by the use of crutches.

Our initial goal for testing in disable assist mode was to provide straight line walking assistance. From initial tests, it was determined that the center of mass of the user while wearing the exoskeleton was shifted too far back. In order to reduce the amount of weight behind the user, the actuated hip ab/adduction actuators and the passive hip internal/external rotation was replaced with a rigid link (see Figure 7). Because straight line walking does not require these degrees of freedom, performance was not degraded.

Figure 7: IHMC Mobility Assist Exoskeleton modified for straight line walking. The hip ab/adduction actuators and the hip internal/external passive rotation degrees of freedom have been replaced with a rigid link.

In disable assist mode, the actuators function under position control and move the joints in such a way to generate locomotion. Joint trajectories were generated through a record application, in which an able bodied user walked wearing the exoskeleton in zero assistance control (zero force control) while the position of the joints and the loading conditions of the feet were recorded. Various types of gaits were recorded, including static and dynamic walks, with a range of swing leg ground clearance gaits. One characteristic of these recorded gaits was intentional limitation of toe off force because the IHMC Mobility Assist Exoskeleton has a passive ankle.

The IHMC Mobility Assist Exoskeleton was first tested in disable assist mode with able bodied users with relaxed leg muscles. In playback mode, the playback speed can be adjusted and playback can be paused at any point during the walking gait. For successful playback there needs to be a dynamic collaboration between the user and the exoskeleton. The user must anticipate the next move of the exoskeleton's lower extremities, and use his/her torso to position his/her body to unload the upcoming swing leg during double support phase. The IHMC Mobility Assist Exoskeleton was tested in disable assist mode with four able bodied users with one recorded gait. Within thirty minutes of practice, all of the users were able to walk a short distance at full playback speed and provide balance support with forearm crutches. Testing will continue with disabled users.

The results from the record and playback trials for one of the users is shown in Figure 8 and Figure 9. During playback, the tracking of the actual joint angles matched the desired values so closely that the desired joint angles were omitted. The gait used for this record data was specific for disable assistance walking. The gait featured small steps, very little knee flexion during support, and large knee flexion during swing.

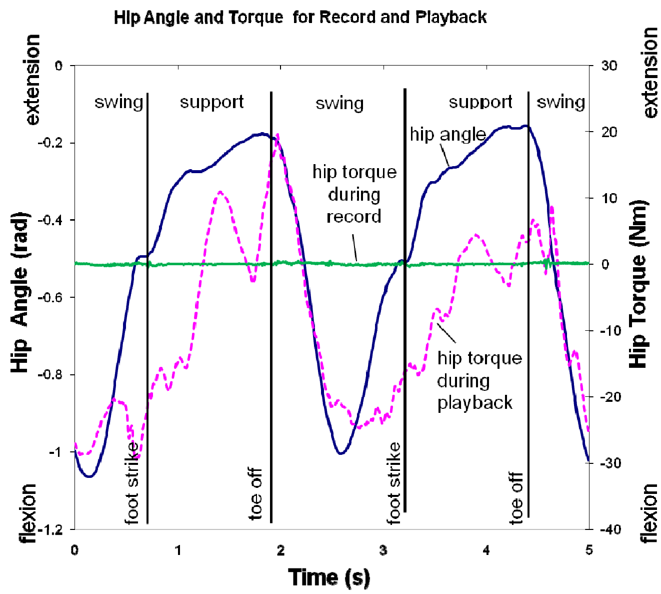

Figure 8: Graph of hip flexion angle and torque data for record and playback while of able bodied user wearing the IHMC Mobility Assist Exoskeleton. During record, the user was intentionally walking with a hyper extended support knee and exaggerated knee flexion during swing for large ground clearance. During record both hip and knee joint torques are approximately zero because the actuators were in zero assistance mode. During playback, the user tried not to use his leg muscles, which is shown by the required joint torque being non-zero.

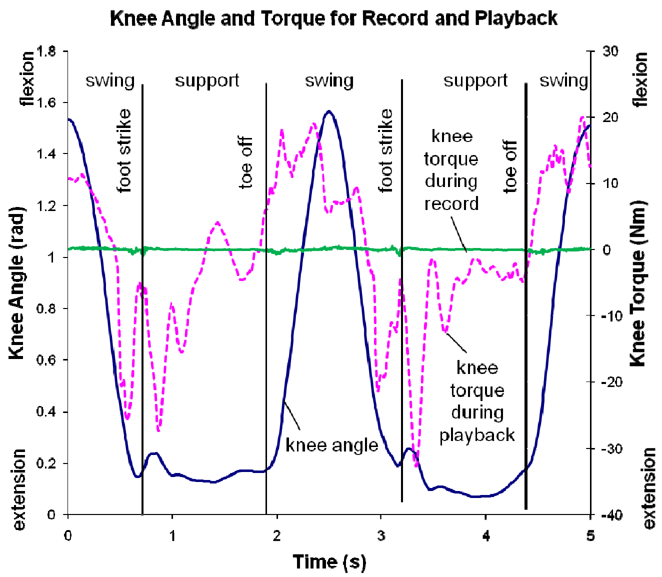

Figure 9: Graph of knee flexion angle and torque data for record and playback while of able bodied user wearing the IHMC Mobility Assist Exoskeleton. During record, the user was intentionally walking with a hyper extended support knee and exaggerated knee flexion during swing for large ground clearance. During record both hip and knee joint torques are approximately zero because the actuators were in zero assistance mode. During playback, the user tried not to use his leg muscles, which is shown by the required joint torque being non-zero.

VIII. Conclusion

In this paper, the design of the IHMC Mobility Assist Exoskeleton, an anthropomorphic lower extremity exoskeleton that can shadow the movement of a healthy pilot and present minimal hindrance to their movement while walking, is presented. The anthropomorphic design is achieved by allowing numerous degrees of freedom for size adjustments, and by the joint design such that the axes of rotation on the exoskeleton intersect with the user's leg and hip joints. For the initial walking algorithm, a lookup table for joint trajectories recorded from the gait of a healthy individual was used, and a position based force control system was implemented to get the exoskeleton to follow the desired joint trajectories with a paralyzed person wearing it. The IHMC Mobility Assist Exoskeleton demonstrated the potential to generate gait on four healthy individuals. Future work will focus on testing the device on people with lower extremity paralysis and further development.

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