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Design and Control of a Robotic Lower Extremity Exoskeleton for Gait Rehabilitation

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Abstract— **Design and control of an active knee rehabilitation orthotic system called ANdROS that was designed as a wearable and portable gait rehabilitation tool is presented. A corrective force field that reinforces a desired gait pattern is applied to the patient's impaired leg around the knee joint via an impedance controlled exoskeleton. The impedance controller is synchronized with the patient's walking phase which is estimated from the kinematic measurements of the healthy leg. The performance of the controller is evaluated through bench-testing.**

I. Introduction

During the past decade, the field of rehabilitation has witnessed an increasing interest for the clinical use of robotic systems; particularly in the treatment of neurological ailments such as stroke and traumatic brain injury. Stroke survivors typically receive intensive, hands-on physical and occupational therapy to encourage motor recovery. Manual treadmill locomotor training with partial body weight support (BWS) approach has been proven effective in improving gait of post-stroke patients [1]. Considering the physical effort involved in such exercises where therapists continually guide the legs and the torso of the patient, robotic neurorehabilitation devices present a great potential as an assistive tool for clinicians by reducing their physical burden. Indeed, some of these systems have already been adopted in clinical practice [2-4]. Other advantages of robotic systems when compared to manual physical therapy include higher precision and repeatability, and quantitative monitoring of patient's progress via sensors [5]. These factors result in faster and greater level of functional recovery [6], thus leading to an improvement in patient's level of independence and quality of life.

According to the National Stroke Association, each year approximately 795,000 people experience a stroke in the United States alone [7] and approximately two-thirds of these individuals survive and require rehabilitation. Impairments such as spasticity, muscle weakness, loss of range of motion (RoM), and impaired force generation create mobility limitations that hinder the activities of daily living (ADL). Only 50-70% of stroke survivors regain functional independence, whereas 15-30% are permanently disabled, and 20% require institutional care at 3 months after onset [8]. Moreover, due to economic pressures on national care systems as well as increasing numbers of cases, patients are receiving less therapy overall and are discharged from rehabilitation hospitals sooner. These statistics suggest that the potential benefits of robotic neurorehabilitation for the individuals as well as for the society are significant.

Ambulatory stroke survivors display substantial alterations in their gait patterns as a result of the compensatory strategies developed in response to their impaired motor control skills. Compromised motor control and force generation frequently lead to limited knee flexion during stance and swing (i.e. stiff-legged gait). Often, knee hyperextension develops in these patients as a mechanism to increase stability during stance. Unfortunately, knee hyperextension during stance may lead to premature degenerative joint disease and therefore clinicians devote part of their effort with post-stroke subjects to seek alternative ways to achieve stability during ambulation. Conventional treatments largely focus on the use of ankle foot orthoses (AFO's) and/or knee orthoses that increase stability by limiting the RoM (thus instigating an abnormal gait pattern). Consequently, the impact of knee orthoses on mobility is rather negative and their use is limited to clinical cases in which the need for providing stability has higher priority than restoring desirable movement patterns. On the other hand, robotic knee orthoses have the potential to overcome the aforementioned limitations by facilitating the knee movement instead of restricting it.

Several robotic devices for gait retraining of stroke patients have been developed in the last decade. The Lokomat (Hocoma AG, Switzerland) [2] is a bilateral gait rehabilitation robot with a BWS system. The patient's legs are actuated in the sagittal plane via DC motors coupled to ball screws. In contrast to Lokomat, new breed of exoskeletons are generally designed to inherit low mechanical impedance. For example LOPES (University of Twente in the Netherlands) [9] uses Bowden-cable driven series elastic actuators. The aforementioned devices control

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multiple degrees of freedom (DoF) of the patient and can only be used in a hospital setting due to their complex design. Even though there exists active lower-extremity exoskeletons that are portable, most are designed to serve as assistive devices for ADL and not for gait retraining. eLEGS from Berkeley Bionics [10] is such a device that enables paraplegics to stand up and walk with the use of a gesture-based human-machine interface, albeit with help of crutches.

In our previous work, we had developed a semi-active brace with controllable damping at the knee joint, achieved through an Electro-Rheological Fluid (ERF) based smart damper [11]. The use of the semi-active ERF damper allowed a compact and lightweight design with highly tunable resistive torque capabilities. However, since the ERF damper is a resistive element, its use for gait retraining was limited to patients with mild to moderate gait deviations. In this paper, we present the design and impedance control of a new active knee rehabilitation orthotic system called ANdROS that can not only apply resistive torques but also generate active corrective torque fields at the knee joint (Fig. 1). It should be noted that ANdROS is designed such that it could also be used in overground walking by migrating the control algorithm to an onboard microcontroller with a battery back (not discussed herein since it is outside the scope of this paper). The allowance of hip abduction/adduction and flexion/extension allows the user to maintain his/her RoM in these planes.

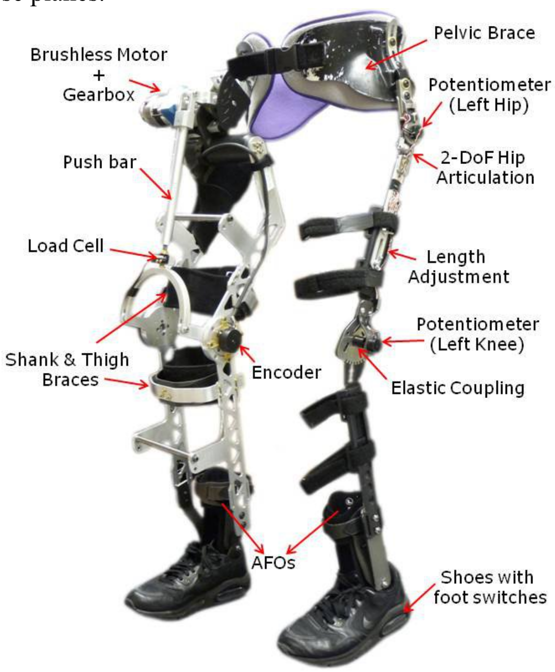

Fig. 1. Close-up view of the new Active Knee Rehabilitation Orthotic System (ANdROS) and its major components. ANdROS has the potential to replace high-cost robotic systems for several patient populations.

Although existing robotic devices provide a valuable asset for rehabilitation hospitals, their high cost limits the number of training sessions the patients receive during rehabilitation. Thus there exists a technological gap for a new breed of rehabilitative orthotic devices that maintain the positive attributes of the treadmill devices while downplaying their high cost. Our long term goal is that ANdROS becomes a low cost, wearable and portable smart gait retraining exoskeleton that could be used by stroke patients at their home setting with constant reinforcement of the targeted gait pattern.

II. System Overview

The lower extremity exoskeleton ANdROS is a wearable and portable assistive tool for gait rehabilitation and monitoring of people with motor control deficits due to a neurological ailment, such as stroke. ANdROS reinforces a desired gait pattern by continually applying a corrective torque around the knee joint, commanded by the impedance controller. A sensorized yet unactuated brace worn on the unimpaired leg is used to synchronize the playback of the desired trajectory based on the user's intent. The device is mechanically grounded through two AFOs rigidly attached to the main structure, which helps reduce the weight perceived by the user.

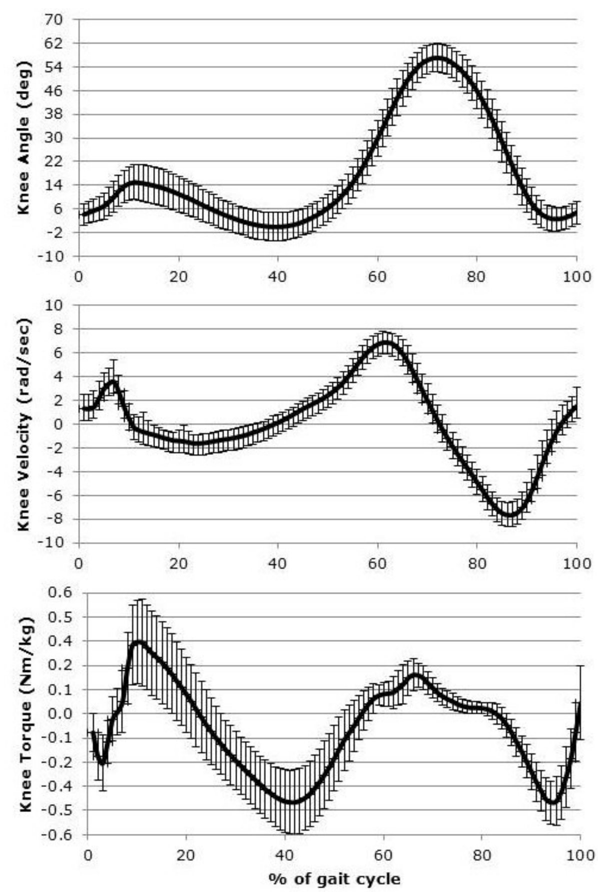

Fig. 2. Gait trajectories obtained from 70 healty subjects using the Vicon motion capture sytem. Average of knee angle (TOP), knee velocity (MIDDLE), and knee torque (BOTTOM) measurements are presented along with minimum and maximum values.

A. Design Requirements

The power and torque specifications of ANdROS were determined based on inverse dynamics and reference trajectories obtained from 70 healthy subjects using the Vicon motion capture system at Spaulding Rehabilitation Hospital in Boston, MA. The results of the tests conducted at an average walking speed of 1.3 m/s are presented in Fig. 2 with the mean, maximum, minimum values. The torque measurements are normalized for the weight of the individual, and power is derived from multiplication of knee velocity with knee torque. Table 1 summarizes the design requirements obtained from these tests, calculated for a 100 kg individual.

TABLE I. DESIGN REQUIREMENTS

Peak Velocity (RPM)	Peak Torque[a] (Nm)	Peak Power[a] (W)
73	44	187

a The torque values are calculated for a 100 kg individual at 1.3 m/s walking speed

B. Design and Operation

The exoskeleton frame consists of two separate leg braces: an actuated brace attached to the impaired leg via rigid straps, and a sensorized brace attached to the healthy leg via elastic straps. The actuated brace is driven by a brushless DC motor (MCG IB23000-E1) coupled to a gearbox (Anaheim Automation GBPH-0602-NP-040, 40:1 gear ratio) which is located next to the pelvis, to reduce the moving mass and keep the center of mass (CM) of the exoskeleton close to the wearer's CM. The motor has been selected to provide at least 50% assistance to a 100 kg individual. The peak torque at the gearbox output is 22.88 Nm and rated power is 179 W. The torque generated by the motor is transferred to the knee joint by means of a push bar that features a quick disengagement mechanism. By disengaging the lower part of the exoskeleton from the power train, easy adjustment during donning/doffing is possible. The push bar mechanism of ANdROS is basically a four bar linkage. Simple kinematic analysis has been carried out to convert the torque at the motor shaft to the torque at the knee, but has been excluded herein for the sake of clarity.

A load cell at the end of the push bar measures the interaction forces. The load cell is located proximal to the knee joint to maximize the amount of mass between the motor and itself, whose apparent inertia is reduced by force-feedback. An incremental encoder and a rotary potentiometer are placed on either side of the knee joint, where the potentiometer is used to initialize the incremental encoder's position. The rotary encoder provides high accuracy (0.088°) and noise free measurement of the knee angle, which are critical properties in calculating digital derivative for velocity estimation.

An exoskeleton should be attached to the body so that there is minimal play (migration) with respect to the human tissue, yet still be comfortable to wear. The actuated brace of ANdROS is attached to the user's body at four locations (i.e. hip, thigh, shank, and ankle) to prevent the migration effect. The weight of the exoskeleton and the user is transferred to the ground through an AFO. The symmetrical structure of the actuated brace applies torque to the knee from both medial and lateral sides of the leg, thus eliminating the twisting effect which would occur in a single-sided design.

Redundant safety features are implemented both at the hardware and software level. The potentiometer that initializes the rotary encoder also serves as a safety measure by comparing its reading by that of the encoder, and halting the program in case of a discrepancy (e.g. due to a sensor failure). Mechanical stops located at the gearbox output shaft prevent the patient from hyper-extending their leg by limiting the RoM, which can be adjusted via screws. A redundant RoM check on the software side sets all outputs to zero and terminates the program in case of violation of the allowed range. A spring loaded stop switch held by the therapist immediately cuts the power to the motor when released. Safety covers are placed around pinching points to avoid injury (not shown in figure).

The brace worn on the unimpaired leg measures the knee and hip flexion/extension angles via high precision potentiometers. The data acquired from these sensors are used in synchronization of the robot with the user. Foot switches are embedded inside the sole of the shoes to be used as a distinct reference point in gait analysis, and to confirm the accuracy of human-machine synchronization. Patient-cooperative strategies are further discussed in Sec. 3.2. The 2-DoF hip articulation also allows for abduction/adduction and flexion/extension of the hip joint, which grants the user maintain their RoM in these planes. It is crucial that in any joint, the center of rotation of the exoskeleton coincides with the wearer's joints to prevent unintended moments and forces on the limbs. Therefore, the thigh and shank components are slotted for adjusting the distance from hip to the knee joint and from the knee joint to the ankle joint, respectively.

C. Control Hardware and Software

The control hardware of ANdROS consists of two personal computers: a host computer, and a real-time target. The host computer runs LabVIEW with a graphical user interface (GUI) through which the practitioner can modify the exercise parameters. Communication between the host and the real-time target is carried out by the LabVIEW Shared Variable Engine (Fig. 3). The real-time target is the dedicated controller of the system that runs all the time critical tasks such as data acquisition (DAQ) and controls. Regular DAQ hardware running on a general-purpose operating system (OS) such as Windows cannot guarantee real-time performance. In contrast, real-time hardware running on real-time OS allows the programmer to prioritize tasks so that the most critical task can always take control of the processor when needed. The real-time target is equipped with a National Instruments 6259 M-Series I/O card. The digital controller implemented on the real-time target operates at a frequency of 10 kHz.

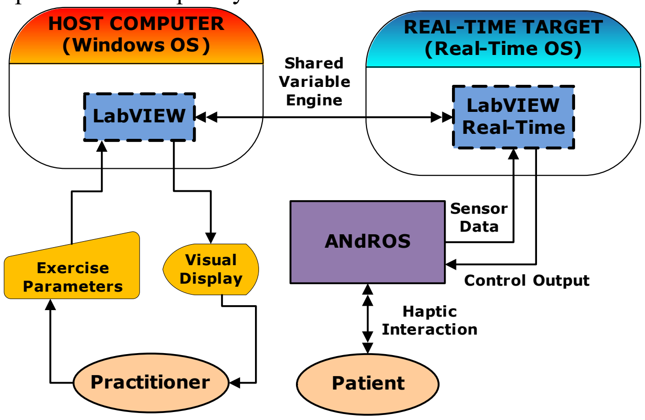

Fig. 3. Framework of the data flow of ANdROS' control system

Fig. 4. Treadmill training with ANdROS

D. *Exercise Modes*

A rehabilitation exercise (robotic or traditional) may be carried out in one of the three ways [12]:

- Passive, where the movement is externally imposed while the patient remains relaxed. Maintaining the range of motion at joints and flexibility in muscle and connective tissue is the main outcome of this type of exercise.
- Active assisted, where a desired movement is manually completed for the patient if they are unable to complete it on their own.
- Active resisted, where the patient must move against a resistance; therefore it is intended for higher level patients.

ANdROS is controlled in the active assisted mode, where the device applies corrective torque to the knee joint of the impaired leg, thus reinforcing a desired gait trajectory while the patient is exercising on a treadmill (Fig. 4). The control algorithm is designed such that instead of imposing a fixed gait trajectory on the patient, the patient is allowed to influence the cadence by the measurement of flexion/extension angles of the hips and the knees (see Section III).

III. Feedback Control

ANdROS targets to reinforce a desired gait pattern by applying a corrective torque field to the wearer's knee using an impedance controller based on the deviation from a reference trajectory.

A. *Impedance Controller*

Mechanical impedance (or simply impedance) is defined as the reaction of a system to an imposed motion. It is the dynamic generalization of stiffness and is a complex function that defines the dynamic behavior of the robot independently from the environment (unlike motion and force). Impedance controlled systems produce a force in response to an imposed motion.

In control of robots that interact with an environment (e.g. humans in case of robotic rehabilitation), the environment is considered an admittance; therefore the robot is controlled to behave as an impedance [13]. One major advantage of impedance control is stable interaction with stiff environments. Fig. 5 shows an example of such an interaction where ANdROS undergoes a stable collision with a rigid environment (the horizontal plateau in the measured position shows that its movement is restricted).

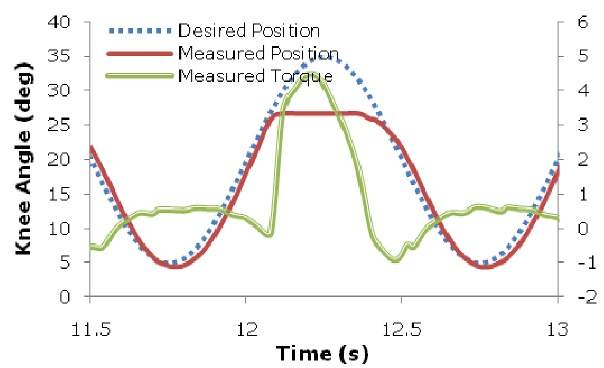

Fig. 5. ANdROS experiencing a stable collision with a rigid environment

Results from the impedance control algorithm are shown in Fig. 6 while the diagram of ANdROS' controller is shown in Fig. 7. The desired knee angle is generated from the previously collected gait trajectories (see Section II.A), and the estimation of gait phase based on the human-machine synchronization algorithm. The synchronization algorithm is a variation of the method proposed by Aoyagi et al. [14]; and not described in details here due to space limitations. The outer position loop simulates the impedance of a spring damper system, and the inner force loop reduces the apparent inertia of the system by using force feedback, as well as compensating for gravitational forces.

The corrective torque applied by the impedance controller can be observed more easily if the reference trajectory is set to zero and the device is externally forced to deviate from the reference. The corrective torque is applied as if there is a virtual spring and a damper attached to the reference trajectory (Fig. 6).

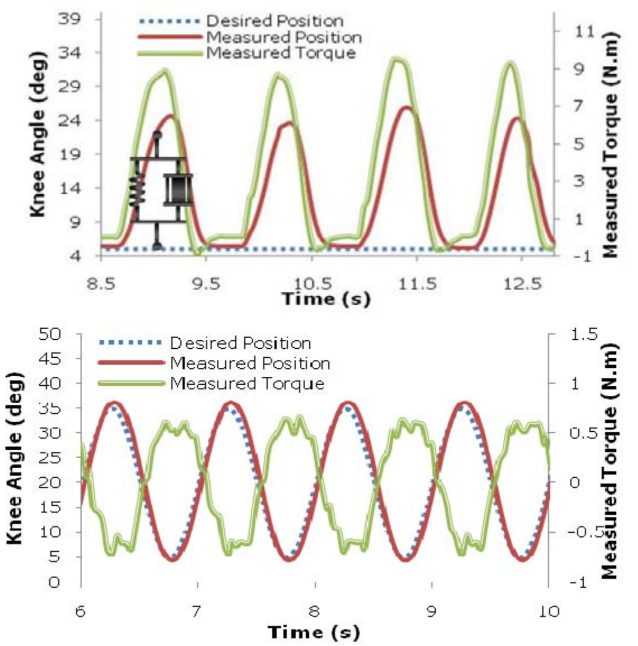

Fig. 6. The impedance controller applies corrective torque as the actual position deviates from the reference; simulating a virtual spring damper system attached to the reference trajectory (TOP); The controller following a 1 Hz sinusoidal trajectory. (BOTTOM).

B. Backdrivability Testing

Backdrivability is defined as the intrinsic low endpoint mechanical impedance. In robots interacting with humans it is desirable that the robot's weight and inertia are imperceptible by the user. When the desired torque is set to zero in the presence of a force-feedback loop (i.e. backdrivable mode), the robot follows the user's motions in an effort to minimize the interaction torque. The effect of force feedback in the inner feedback loop is to increase backdrivability by reducing the apparent inertia of the system [15].

The weight of the lower frame and the shank of the user creates a torque around the knee. To compensate for gravity, a counter-torque which is a function of knee angle is applied by the motor (assuming that the subject is standing upright). However, when gravity compensation is present the load cell also measures the corrective force applied by the motor. To isolate these forces and render the system sensitive to external forces only (i.e. to those applied by the user) the gravitational force is subtracted from the measured force (Fig. 7).

The actual interaction forces between the user and the exoskeleton cannot be measured by the load cell that is located on the push-bar, since the forces arising from inertia, weight, and friction of the lower brace are also sensed at this location. Therefore, a secondary load cell is attached to the lower brace during backdrivability testing to estimate the forces that would be perceived by the user at their shank (Fig.8-TOP). The effect of gravity compensation and force-feedback on interaction forces is shown in Fig.8-BOTTOM while the exoskeleton is being oscillated manually when the desired torque is set to zero. It can be observed from the graph that gravity compensation reduces the interaction forces by offsetting it towards zero, whereas backdrivability reduces the interaction forces by decreasing the amplitude as a result of reduced apparent inertia.

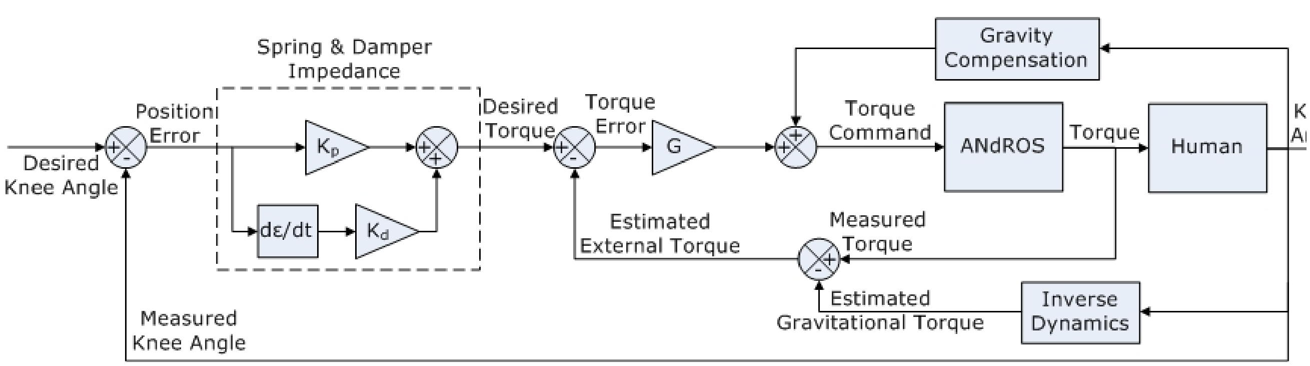

Fig. 7. The block diagram of the impedance controller implemented on ANdROS.

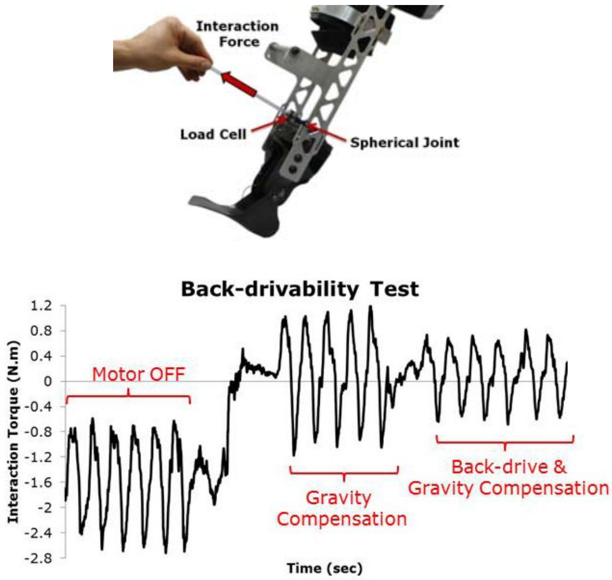

Fig. 8. The test setup for measuring the actual interaction forces between the human and the exoskeleton (TOP); Comparison of interaction torque experienced by the user (BOTTOM). Gravity compensation offsets the torque, whereas backdriving reduces the amplitude of torque

IV. Conclusion

A robotic lower-extremity exoskeleton designed as a wearable and portable assistive tool for gait rehabilitation and monitoring was developed. The design and functionality of the system was discussed, and results on the impedance controller were presented. The impedance control algorithm provides stable interaction by defining the dynamic relationship between the human and the exoskeleton. The implemented controller simulates the impedance of a virtual spring and damper system. The inner force feedback control loop reduces the apparent inertial of the system. Together with the gravity compensation algorithm the interaction forces are reduced, rendering the exoskeleton transparent to the user. A recent trend in robotic rehabilitation is the application of patient-cooperative exercise regimens, where the robot adapts its behavior based on the patient's voluntary participation [2]. In other words, patient can influence the cadence. In robot assisted gait therapy, the reference trajectory of the impedance controller is generally a fixed trajectory that is replayed at a constant period. As indicated by Aoyagi et al. [14], this approach results in a phenomenon similar to that of a pair of coupled oscillators where the human starts walking out of phase with the robot. To avoid this phenomenon, a human-machine synchronization algorithm that is based on the work of Aoyagi et al. [14] is implemented on ANdROS. The algorithm uses flexion/extension angle measurements from both knees and hips to estimate the gait phase and sets the corresponding desired knee angle from the reference trajectory. The foot switches embedded in the sole of the shoes are only used to verify the accuracy of the algorithm, and not used in the algorithm itself due to robustness concerns.

Future work includes implementation of friction compensation algorithm and improvement of human-machine synchronization, followed by healthy subject testing on a treadmill. The final goal of this research is to provide an affordable gait rehabilitation tool for the health care industry.

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