Ph.D. DISSERTATION

Design, Implementation and Control of an Overground Gait and Balance Trainer with an Active Pelvis-Hip Exoskeleton

by

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DESIGN, IMPLEMENTATION AND CONTROL OF A ROBOT-ASSISTED GAIT TRAINER WITH AN ACTIVE PELVIS-HIP EXOSKELETON

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Abstract

Human locomotion is crucial for performing activities of daily living and any disability in gait causes a significant decrease in the quality of life. Gait rehabilitation therapy is imperative to improve adverse effects caused by such disabilities. Gait therapies are known to be more effective when they are intense, repetitive, and allow for active involvement of patients. Robotic devices excel in performing repetitive gait rehabilitation therapies as they can eliminate the physical burden of the therapist, enable safe and versatile training with increased intensity, while allowing quantitative measurements of patient progress. Gait therapies need to be applied to specific joints of patients such that the joints work in a coordinated and repetitious sequence to generate a natural gait pattern. Six determinants of gait pattern have been identified that lead to efficient locomotion and any irregularities in these determinants result in pathological gaits. Three of these six basic gait determinants include movements of the pelvic joint; therefore, an effective gait rehabilitation robot is expected to be capable of controlling the movements of the human pelvis.

We present the design, implementation, control, and experimental verification of AssistOn-GAIT,a robot-assisted trainer, for restoration and improvement of gait and balance of patients with disabilities affecting their lower extremities. In addition to overground gait and balance training, AssistOn-GAIT can deliver pelvis-hip exercises aimed to correct compensatory movements arising from abnormal gait patterns, extending the type of therapies that can be administered using lower extremity exoskeletons.

AssistOn-GAIT features a modular design, consisting of an impedance controlled, self-aligning pelvis-hip exoskeleton, supported by a motion controlled holonomic mobile platform and a series-elastic body weight support system. The pelvis-hip exoskeleton possesses 7 active degrees of freedom to independently control the rotation of the each hip in the sagittal plane along with the pelvic rotation, the pelvic tilt, lateral pelvic displacement, and the pelvic displacements in the sagittal plane. The series elastic body weight support system can provide dynamic unloading to support a percentage of a patient's weight, while also compensating for the

inertial forces caused by the vertical movements of the body. The holonomic mobile base can track the movements of patients on flat surfaces, allowing patients to walk naturally, start/stop motion, vary their speed, sidestep to maintain balance, and turn to change their walking direction. Each of these modules can be used independently or in combination with each other, to provide different configurations for overground and treadmill based training with and without dynamic body weight support.

The pelvis-hip exoskeleton of AssistOn-GAIT is constructed using two passively backdrivable planar parallel mechanisms connected to the patient with a custom harness, to enable both passive movements and independent active impedance control of the pelvis-hip complex. Furthermore, the exoskeleton is self-aligning; it can automatically adjust the center of rotation of its joint axes, enabling an ideal match between patient's hip rotation axes and the device axes in the sagittal plane. This feature not only guarantees ergonomy and comfort throughout the therapy, but also extends the usable range of motion for the hip joint. Moreover, this feature significantly shortens the setup time required to attach the patient to the exoskeleton. The exoskeleton can also be used to implement virtual constraints to ensure coordination and synchronization between various degrees of freedom of the pelvis-hip complex and to assist patients as-needed for natural gait cycles.

The overall kinematics of AssistOn-GAIT is redundant, as the exoskeleton module spans all the degrees of freedom covered by the mobile platform. Furthermore, the device features dual layer actuation, since the exoskeleton module is designed for force control with good transparency, while the mobile base is designed for motion control to carry the weight of the patient and the exoskeleton. The kinematically redundant dual layer actuation enables the mobile base of the system to be controlled using workspace centering control strategy without the need for any additional sensors, since the patient movements are readily measured by the exoskeleton module. The workspace centering controller ensures that the workspace limits of the exoskeleton module are not reached, decoupling the dynamics of the mobile base from the dynamics of the exoskeleton. Consequently, AssistOn-GAIT possesses virtually unlimited workspace, while featuring the same output impedance and force rendering performance as its exoskeleton module. The mobile platform can also be used to generate virtual fixtures to guide patient movements.

The ergonomy and useability of AssistOn-GAIT have been tested with several human subject experiments. The experimental results verify that AssistOn-GAIT can achieve the desired level of ergonomy and passive backdrivability, as the gait patterns with the device in zero impedance mode are shown not to significantly deviate from the natural gait of the subjects. Furthermore, virtual constraints and force-feedback assistance provided by AssistOn-GAIT have been shown to be adequate to ensure repeatability of desired corrective gait patterns.

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Chapter I

1 Introduction

Human well being and social interaction are directly related to daily activities performed by the body, the most important among which is walking. Therefore, a disability in walking has a profound impact on the quality of life [4, 5]. Disabilities of joints used in gait can be caused by a large number of factors which include but are not limited to injury, stroke and nerve disorders. After such a disability is encountered, the patients tends to adapt abnormal gait patterns in order to compensate for the impairment. This can lead to changes in stride length, walking speeds and joint asymmetries. Often, the stronger limb has to bear larger than normal loads which may cause pains or degeneration of the bones [6].

Significant efforts are needed to restore normal gait fully or partially and enable the patient to perform daily tasks independently. These efforts include intense and repetitive rehabilitation therapy. Such therapy is traditionally conducted by trained human therapists, however, the length and effectiveness of the therapy is limited by the therapists' ability to sustain long sessions. Due to recent technological advancements robots are being increasingly used for gait assistance and rehabilitation therapy, as they are suitable for repetitive jobs and also allow for generation of different scenarios with performance measurement.

1.1 Importance of Pelvic Movements during Walking

Gait is the movement of limbs in a certain pattern for moving the whole body over a solid surface. Humans, being bipedal move their lower body joints in a coordinated and harmonious manner while the muscles and momentum drives the motion of the body Center of Gravity (CoG) and the trunk. Specific joint movements that are crucial for gait were determined by [7]. These are the *pelvic rotation* in the transverse plane, *pelvic tilt* in the coronal plane, the knee and hip flexion, the ankle plantar flexion, the foot and ankle rotations and the *lateral pelvic displacement* (Figure 1.2). These determinants minimize the movement of the body CoG to ensure efficient locomotion. Any irregularities in one of these determinants will cause the gait to deviate from normal and be termed as pathological.

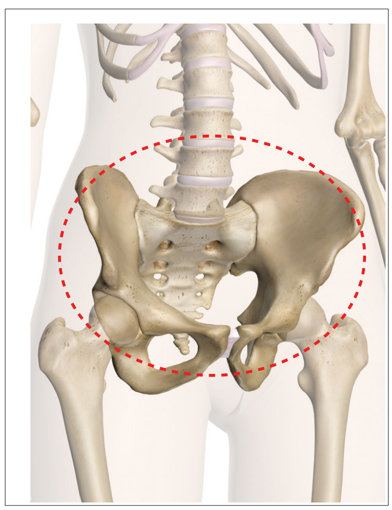

Figure 1.1: The human pelvis

The human pelvis (Figure 1.1) is a collection of bones that supports the spine and is itself supported by the lower limbs. Three out of the six basic determinants of gait include pelvic movements in them which emphasizes the importance of this joint for gait [7]. The pelvic rotation, pelvic tilt, vertical and lateral pelvic displacements are the degrees of freedom (DoF) of the pelvis that are crucial for proper balance and efficient locomotion. In particular, during normal walking, the pelvis rotates about 4° towards the non-load bearing side about the transverse axis to push the hip forward. Known as the *pelvic rotation*, this motion plays an important role in the determination of the stride length and the walking speed, while also reducing the distance traveled by the body CoG over the foot that is bearing the body weight. Another important determinant of the process is the *pelvic tilt* in the coronal plane, which minimizes the displacement of the body CoG during normal walking by reducing the distance between the swinging foot and the ground, by shifting the load bearing foot 5° downwards in the frontal plane. These movements are applied in harmony with knee and ankle flexion in order to establish sufficient ground clearance for the swinging leg. Starting with the initial touch to the ground and finishing at the end of the stance phase, the ankle plantar flexion and foot supination are other important factors that reduce the sinking of the body CoG by preventing the reduction of leg length during walking. The pelvic rotation and pelvic tilt are crucial movements that keep the displacement of the body CoG in the frontal plane around 50 mm during normal walking, while the *lateral pelvic displacement* reduces the shifting of the body center of gravity in the transverse plane by sliding the pelvis over the load bearing leg and plays an important role in maintaining balance during walking.

1.2 Pelvic Movements for Gait Rehabilitation

The pelvis is responsible for transferring the forces of the lower extremity, assisting in forward propulsion, keeping the body CoG over the feet during swinging motion of the legs and modulating the vertical motion of the CoG to reduce energy consumption during ambulation. Therefore, administering proper movements of the pelvis during physical therapies is known to lead to significant improvements in the gait of patients [8, 9]. Widely accepted rehabilitation techniques, such as Bobath and proprioceptive neuromuscular facilitation, emphasize the importance of pelvis movements in the walking cycle [10–12]. Consequently, during conventional Body Weight Supported Treadmill Training (BWSTT), a trained physiotherapist is employed only for managing pelvis movements. Pelvis movements have largely been ignored in robot assisted therapies, which primarily focus on providing appropriate leg movements during walking.

Compensatory movements arising from pelvis motor control problems are the most common abnormal gait deviations. Overall, the most common gait deviation is the hemiplegic gait. Deviation patterns are mostly related to pelvis elevation and hip circumduction. Stroke patients exhibiting hemiplegic gait, for example raise their hemiparetic side abnormally for achieving adequate leg room in the swing phase, causing excessive pelvic elevation. Weakness of distal muscles results in excessive pelvic rotation coupled with an exaggerated hip rotation, causing hip circumduction [13, 14].

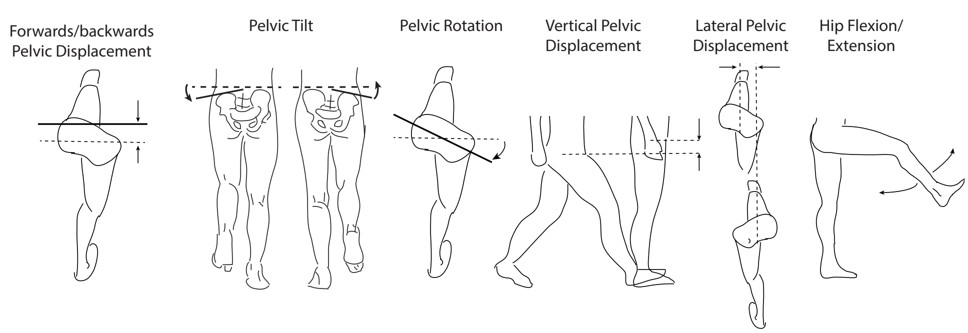

Figure 1.2: Pelvis-Hip movements during walking

1.3 Overview of AssistOn-GAIT

This dissertation presents design, implementation, control and experimental verification of AssistOn-GAIT [15], a robot-assisted overground/treadmill based gait and balance training device, designed to administer therapeutic gait training and balance exercises to patients who have suffered injuries that affect the function of their lower extremities. The device focuses on gait deviations caused by compensatory pelvic movements and features guidance of these movements to restore normal gait through active actuation and control of 7 DoF of the pelvis-hip complex. As shown in Figure 1.3, Figure 3.3 and 3.4 the device consists of a mobile base/treadmill, a pelvis-hip exoskeleton (EXO) [including lateral pelvic actuation mechanism (LPAM)] and an active body weight support (BWS) system. The device can be used with a treadmill or in an overground configuration. Detailed breakdown of the treadmill version is shown in Figure 1.3.

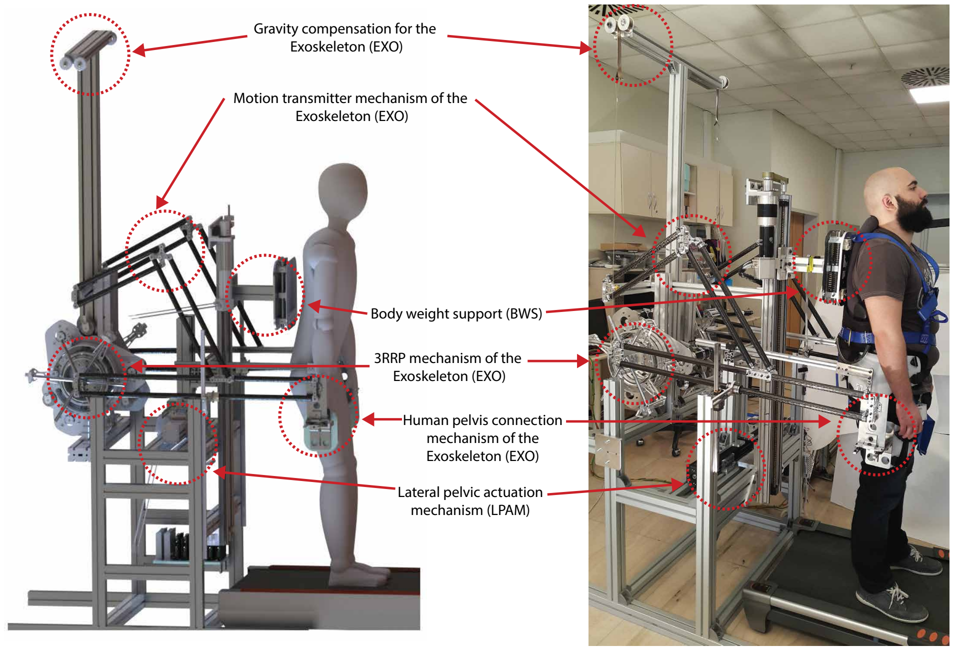

Figure 1.3: (L) A CAD rendering of AssistOn-GAIT showing all modules (R) Prototype of AssistOn-GAIT attached to a volunteer

1.4 Contributions of the Dissertation

The contributions of this dissertation can be summarized as follows:

- AssistOn-GAIT, a novel Robot Assisted Gait Training (RAGT) device has been designed, implemented, controlled and experimentally verified. The device can administer both overground and treadmill based gait and balance rehabilitation training to adult patients (supported hip width 270-420 mm, supported hip height 653-1202 mm, supported weight upto 120 kg) with gait and balance disorders, while actively supporting their weight. Capabilities of the device are specially suited for correction of compensatory pelvic movements that arise from abnormal gait patterns adapted by patients as a result of injury or stroke.

- The device is different from the existing devices in literature in that it allows for and can actively control all degrees of freedom of the pelvis-hip complex that are crucial for gait, while the connected patient walks naturally on the ground/treadmill. Existing devices either restrict pelvic movements or do not allow for natural walking on the ground. Furthermore, the device allows patients to freely swing their arms, a feature that is important for maintaining balance during walking, which is neglected in most of the existing devices. The device also features an active body weight support system capable to providing dynamic weight support while walking. Furthermore, body weight support system is capable of feed forward inertia compensation, ensuring a walking feel that is similar to walking in a low gravity environment. Inertia compensation is a novel feature that is not available in any other body weight support system presented in the literature.

- **Design:** AssistOn-GAIT has a modular design with three modules: holonomic mobile base (BASE), a self aligning exoskeleton (EXO) and an active body weight support (BWS) system. These modules can be operated independently or in combination with each other such that the device can be used in different configurations. When EXO and BWS are mounted on a static platform with a treadmill, they form a treadmill based rehabilitation device. When EXO and BWS are mounted on BASE, they form an overground training device that can also deliver balance excercises in addition to gait rehabilitation. When the BWS is mounted on a static platform with a treadmill, a treadmill based gait trainer with active weight support can be formed. Similarly, a combination of BWS with BASE forms an overground gait trainer with active weight support. The modules of AssistOn-GAIT are designed as follows:

- *Holonomic Mobile Base (BASE)* is designed as a mobile platform that carries the weight of the whole hardware and patients without reflecting its inertial dynamics while following patients transparently to allow them to move forwards/backwards, sidestep and stop as desired.
- *Pelvis-Hip Exoskeleton (EXO)* is designed to actuate 7 degrees of freedom (DoF) of the pelvis hip complex. These movements are rotation of the each hip in the sagittal plane along with the pelvic rotation, the pelvic tilt, lateral pelvic displacement, and the pelvic displacements in the sagittal plane. In line with this, EXO consists of two (left and right) planar mechanisms each providing 3 DoF in the sagittal plane. Additionaly, EXO has a Lateral Pelvic Actuation Mechanism (LPAM) that assists the lateral pelvic movement. Custom motion transmission and human connection mechanisms have been designed that transmit all EXO movements to the human pelvis-hip complex, in a way that the human is placed away from the complex machinery and is also able to freely swing his/her arms.
- *Active Body Weight Support (BWS) System* is designed to dynamically support the weight of the patient while walking. It consists of a series elastic actuator that can move along the longitudinal

(up-down) axis in a way to maintain a constant amount of interaction force with the patient such that a predetermined percentage of the patients' weight actively supported.

The human connects to BWS by a harness that enables unimpeded natural walking, while the weight is supported. Furthermore, BWS features an acceleration sensor that is used for feed forward compensation of inertial forces. The system has been designed to have a sufficiently long stroke such that it can also lift a patient from a sitting position and guarantee safety in case a patient falls while attached to the device.

- **Control:** AssistOn-GAIT can control the interaction forces between patients and the device, and can provide haptic guidance to assist the patients as-needed to help correct compensatory gait patterns. High fidelity force control of the overall device is achieved by employing appropriate control techniques for each module.

  - *BASE* is motion controlled and features a motion control bandwidth that is significantly faster than that is required by patients during walking. This enables motion controlled BASE to react fast enough to track human movements. Two types of control architectures have been implemented for BASE. First is based on series elastic actuation that relies on sensing interaction forces between BASE and the patient. This closed loop force control scheme allows virtual fixtures to be generated for guiding patients along a predefined path on the ground, while patients can move forwards/backwards/sideways.

The second control scheme applied to BASE is based on workspace centering control that relies on displacement measurements by EXO to motion control BASE in such a way that the patient cannot reach the workspace limits of EXO. Given the full kinematic redundancy of the mobile base with respect to the exoskeleton, workspace centering control technique can be applied to control to holonomic base such that the overall device possesses the same force rendering fidelity as its EXO module. Overall, BASE can impose desired trajectories, as well as provide haptic guidance to help patients walk naturally without exceeding safety limits.

- For *EXO*, the left and right planar parallel mechanisms are actuated using direct drive grounded DC motors coupled to a low friction/backlash capstan transmission. This makes these mechanisms passively backdriveable; they feature excellent force control performance even under open loop impedance control. Impedance control of these mechanisms allows generation of virtual fixtures along complex pelvis-hip paths and enable guidance around corrective pelvis-hip trajectories.
- *LPAM part of EXO* allows for support of the lateral pelvic displacements. Series elastic actuation is employed to ensure high fidelity force control within the patient bandwidth. The controller allows for generation of haptic guidance trajectories for the pelvis as necessitated by the rehabilitation protocol.
- *BWS* supports the human weight throughout training and ensures safety under falls. Similar to LPAM, series elastic actuation is employed to ensure high fidelity force control within the patient

bandwidth. The controller allows for maintenance of constant interaction force despite vertical movements of the human body, thus actively supporting a percentage of patients' weight.

  - *BWS* also features compensation for additional vertical forces caused by the vertical accelerations of the human body while walking. This is performed using an emulated inertia compensation scheme that utilizes filtered acceleration measurements (from the acceleration sensor installed on BWS) to estimate inertial forces and provide feed forward force references to approximately compensate for these forces.

- **Device characterization and Human Subjects Experiments** Performance of each module is experimentally characterized and human subject experiments with healthy volunteers are conducted to verify the performance of AssistOn-GAIT .

  - The workspace of the device along the crucial 7 DoF of the pelvis-hip complex has been characterized by physically measuring the amount of translation/rotation that can be performed by a human subject connected to the device. It has been shown that the overall device capabilities satisfy the workspace required by pelvis-hip during walking.
  - Trajectory tracking performances of the EXO, LPAM and BWS have been characterized. It has been shown that each of the modules are able to accurately track desired reference trajectories less than 1.5% RMS error.

- Motion control bandwidth of BASE, force control bandwidth of LPAM and force control bandwidth of BWS have been characterized experimentally. It has been shown that these bandwidths significantly exceed the bandwidth required for human walking.
- Backdriveability of EXO has been experimentally verified and it has been shown that less than 6 N force is required along the translations degrees of freedom and less than 0.4 Nm torque is required along the rotational degree of freedom, to move EXO when in is not actuated.
- Impedance rendering capability of EXO has been verified under open loop impedance control and a rendering error of 3% RMS has been reported.
- Donning time (time required to connect a patient to the device) has been experimentally determined to be less than 5 minutes for the first session and less than 1 minute for subsequent sessions with the same patient.
- Experiments for data collection have been performed with healthy humans. While connected to the device, the human is asked to walk on a treadmill and data related to 7 crucial DoF has been collected. This data collection helps with determining the gait of the human and the path along which haptic guidance needs to be provided. Furthermore, it may help with analysis of abnormalities in gait.
- Adequacy of the haptic guidance provided to the pelvis-hip complex has been experimentally verified with healthy subjects. Vir-

tual fixtures are generated along desired pelvis-hip trajectories based on data collected from earlier walking sessions. It has been shown that wider tunnels allow subjects to walk with self-selected speeds within the tunnels and any large deviation from the desired trajectories result in gentle guidance forces that bring pelvis-hip complex back to within the virtual tunnel. Similarly, narrow tunnels can strictly keep the pelvis-hip complex on the desired path.

- Inertia compensation of BWS has been experimentally verified with healthy subjects walking while connected to BWS. Ground reaction forces estimated with and without intertia compensation indicate that inertia compensation control allows for more natural ground reaction forces to be felt by the user.

- Experiments with a motion tracking system have been conducted to verify that AssistOn-GAIT provides minimal hinderance in natural walking when the patients are connected to it. A comparison of gait patterns with and without AssistOn-GAIT indicate that the device provides sufficient transparency to allow natural walking when patients are connected to it.

1.5 Outline of the Dissertation

This dissertation is arranged as follows:-

- Chapter I : Discussion on the motivation behind the dissertation and an overview of the dissertation.
- Chapter-II : Discussion on existing devices that have been surveyed for the dissertation, their features and shortcomings.
- Chapter-III : Introduction to the proposed research and why this dissertation is important.
- Chapter IV : Discussion on the design and implementation of the Pelvis-Hip Exoskeleton (including the Lateral Pelvic Actuation Mechanism) that provides 7DoF pelvis-hip movements
- Chapter V : Discussion on the Series Elastic Actuation and Workspace Centering Control of the Mobile Base
- Chapter VI : Discussion on Design and Implementation of the Active Body Weight Support System with Inertia Compensation
- Chapter VII : Discussion on the performance characterization of the device and experiments performed on healthy subjects
- Chapter VIII : Concludes the report and Discusses Future Works
- Appendices : Additional information

Chapter II

2 Literature Review

In this chapter we briefly summarize the different categories of gait training robots and review relevant devices that have been presented in literature. The shortcomings of each of the devices are discussed and compared with the capabilities of the proposed device.

2.1 Robot Assisted Gait Training

Traditionally, gait rehabilitation therapy is performed by trained human therapists. Multiple therapists help the patient by supporting body weight and assisting leg, knee and pelvis movements. The most basic form of gait rehabilitation is manual gait rehabilitation in which patients are made to practice and repeat specific exercises. In many cases, patients may not able to support their weight fully, therefore, parallel bars are utilized where patients can use their upper body strength to support their weight. A human therapist may also be dedicated for the purpose of supporting the patients weight. Improvement over the manual procedure is partial Body Weight Supported Treadmill Training (BWSTT) which has been shown to be more effective [16,17]. Conventional BWSTT training commonly necessitates a team of three trained therapists to work together with the patient. One of these therapists helps the patient with proper trunk/pelvis alignment and weight shifting, while the others assist with the leg control during stance and swinging of the limb [18].

Improvements to this technique are the therapist assisted Body Weight Supported Treadmill Training (BWSTT) and therapist assisted Body Weight Supported Overground Training (BWSOT) where the patients weight is partially supported by an overhead harness and human therapists help with joint movements. The efficacy of these techniques is limited by the time human therapists can sustain labor intensive therapy sessions and the inability to accurately ensure joint trajectories. Furthermore, correct assessment of patient performance is not possible. Use of robotic devices for performing rehabilitation therapy can help overcome these drawbacks by easily handling repetitive tasks thus reducing the physical workload of therapists, enabling a therapist to work with more than one patients at the same time and reducing the number of required physiotherapists and the cost of therapy to make it more accessible. A number of features that were previously impossible can now be offered such as quantitative measurement of patient progress while enhancing the reliability, safety and accuracy of treatment. In addition to this, interactive therapy with active patient participation with custom duration and intensity can be administered [19]. Clinical trials of robotic rehabilitation provide evidence that robotic therapy is effective for motor recovery and possesses high potential for improving functional independence of patients [20–29]. RAGT devices aim to provide a rehabilitation experience analogous to trained human therapists but in an automated and more systematic way. Based on how these devices actuate human joints, they can be grouped as *exoskeleton type* and *end effector type* [23]. In addition these devices may be used with a *treadmill or overground* and may or may not feature a *partial body weight support*. Another category of RAGT devices are *wearable exoskeletons* which can be used with a treadmill or overground. These categories are briefly discussed in the paragraphs below.

2.1.1 End Effector Type Devices

End-effector type devices connect to the patient from a single point and movements/forces are applied to the patient only at this point. Some of the available end-effector type gait training devices include LokoHelp [30], Gait Trainer GT1 [31], HapticWalker [32], MIT-Skywalker [33] and NEURO-Bike [34]. The movements of these devices do not conform to human joints and joint specific therapies are not feasible without external restraints. Furthermore, these devices allow for compensatory movements of the patient, which may lead to patients adapting inefficient gait patterns. Therefore, these devices are not explored further in this paper.

2.1.2 Exoskeleton Type Devices

Exoskeleton type devices attach parallel to and move in coordination with specific joints and so they can effectively deliver joint specific therapies. Controlled trajectories, forces and torques can be applied to individual joints, making these devices a true improved replacement of trained human therapists. Advanced features like interactive training and measurement of patient performance is also achievable. Exoskeleton type devices have been presented for treadmill training as well as overground training. Wearable exoskeletons such as the HAL [35], H2 [36] and ReWalk [37] are designed to be light-weight and move with the human. Currently with respect to gait rehabilitation, these devices offer limited features and are therefore not discussed further.

Lokomat presented by [38] is one of the most commercially successful RAGT devices. It is a treadmill based device that can control the rotation of the hips and knees in the sagittal plane to ensure mobilization of the legs in a nominal gait pattern. The optional FreeD module allows active control of lateral pelvic displacement, while other movements of the pelvis are not controlled. The LOPES II [39] is another recent treadmill based exoskeleton type device. The device uses a shadow-leg approach and features eight powered degrees of freedom including hip flexion/extension, hip abduction/adduction, pelvis forward/aft and pelvis lateral. Other degrees of freedom of the pelvis are passively allowed but cannot be controlled. Both of these devices focus on the hip and knee rotations in the sagittal plane, these systems give minimal attention to balance and are ineffective against compensatory movements arising from abnormal gait patterns due to unnatural pelvic movements.

Pelvic Assist Manipulator (PAM) can enable/assist pelvis movements during treadmill training. The system utilizes six pneumatic cylinders to actuate 5 DoF pelvic movements (pelvic tilt in the sagittal plane is passive) during BWSTT therapies [40]. Since the hip rotations are not a part of PAM, synchronizing the natural movement of the pelvis with the lower extremity is challenging and necessitates additional control effort [41]. Robotic Gait Rehabilitation (RGR) system [42] uses two single DoF linear actuators at both sides of the hip in order to partially assist pelvic movements. As the system only actuates movements in the vertical plane, the vertical pelvic displacement and pelvic tilt in the coronal plane can be supported actively, while other pelvic rotations and lateral pelvic displacement cannot be actively controlled. Motorika Reoambulator [43] is a commercial rehabilitation device that allows the patient to practice rehabilitation exercises on a treadmill. The device however constrains abduction, pelvic translations and pelvic rotations. ALEX [44] does not allow the anterior/posterior translation of the pelvis and rotations other than vertical rotation.

2.2 Mobile Devices for Robot Assisted Gait Training

A number of exoskeleton type RAGT devices feature mobile bases that carry the load of hardware and allow the patient to move on the ground. The mobile bases follow and assist the patients' walking movements on ground. This helps patients to experience all the sensory inputs associated with walking and move under their own control. This makes overground devices advantageous for gait and balance training. Furthermore, walking on the treadmill has been demonstrated to be different from natural overground walking. Significant differences have been noted in cadence, stride length, stride angles, moments and power [45]. Analysis of pelvis kinematics during treadmill and overground walking has also revealed differences in pelvic rotation and obliquity [46]. Therefore, with proper functionality, overground RAGT may be able to offer a walking experience that is closer to natural gait. The ability to move under their own control may also increase the level of motivation for patients undergoing therapy sessions.

MOPASS presented by [47] features a mobile base to which the patient is connected. The device can generate custom trajectories for the hip and knee joints. With regards to the pelvis, it only actively controls the pelvic rotation in the transverse plane, and thus would not be effective against compensatory movements of the pelvis. The CPWalker presented by [48] is designed to help children with cerebral palsy and make their hips, legs and ankles move according to desired references. This device however, does not control pelvic movements. KineAssist [49, 50] and Walk Trainer [51, 52] are commercial examples of overground gait trainers. Both devices consist of a mobile base connected to a pelvis orthosis and can provide partial body weight support. In particular, KineAssist is equipped with an admittance controlled Cobot base that compensates for the robot dynamics based on the forces measured at its custom designed torso-pelvis harness, which allows for passive movements of the pelvis. KineAssist complies with the natural user movements during walking but cannot assist pelvis/hip movements to help improve the quality of gait. Another overground trainer that relies on the interaction force measurements between the patient and the mobile base is proposed in [53], where unlike KineAssist, a holonomic mobile base is controlled according to these forces. Walk Trainer features active pelvis and leg exoskeletons attached to a differential-drive mobile base [52]. Thanks to a parallel mechanism actuating all six DoF at the pelvis, the system has the ability to actively assist all the pelvic movements. Walk Trainer detects patient intentions regarding to walking speed and heading utilizing two potentiometers and can control its differential-drive mobile base to follow straight and curved paths. NaTUre-gaits [54] is a similar device, consisting of pelvis and leg modules connected to a differential-drive mobile base. This system employs dual three DoF actuated Cartesian planar robots at each side of the hip to assist pelvic movements. These Cartesian planar robots are also used to measure local pelvic motions and these measurements are mapped to walking speed and heading angle to control the differential-drive mobile base of NaTUre-gaits to follow straight and curved paths. Both WalkTrainer and NaTUre-gaits have relatively complex mechanical designs and possess passively non-backdriveable power transmission that necessitates the use of force sensors and active control algorithms to ensure synchronization between the mobile base, pelvis movements and leg rotations to achieve a natural gait for the patient. Furthermore, featuring differential drive mobile bases, both devices are non-holonomic and cannot allow for lateral movements, such as sidestepping.

The device presented in this dissertation features active control of all crucial degrees of freedom of the pelvis and allows the connected patient to walk freely on the ground in any direction. Details are discussed in the subsequent chapters.

2.3 Body Weight Support in Gait Rehabilitation

Patients undergoing gait rehabilitation are often unable to bear their own weight which necessitates the use of Body weight support (BWS). Clinical studies indicate that the effectiveness of gait rehabilitation can be enhanced when partial weight of the patient is supported by a BWS system [55–57]. In addition to this, BWS enhances safety during training by helping to maintain balance and preventing falls. Lateral balance of patients has also been shown to improve, when BWS is used in gait rehabilitation [58]. With respect to weight unloading, these systems can be categorized into i) static systems, ii) passive counterweight based systems, iii) passive elastic spring based systems, and iv) active dynamic systems. Figure 2.1 presents a schematic representation of these categories [1].

The first three types are passive systems which are unable to provide a constant weight support while the pelvis moves vertically, and are therefore not suitable for gait rehabilitation. *Active dynamic* systems are capable of generating unloading forces dynamically [1]. In particular, these type of

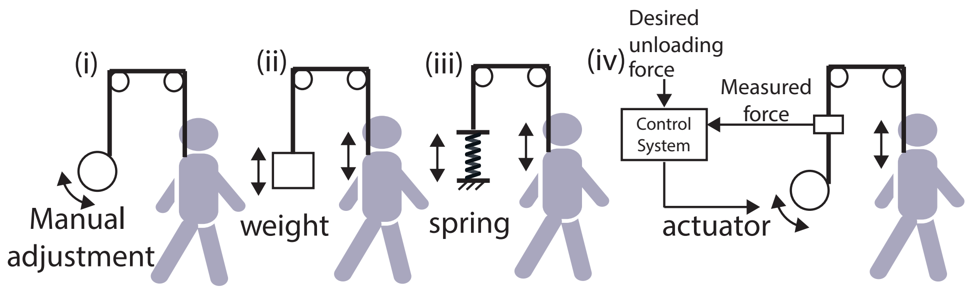

Figure 2.1: Types of BWS systems (i) Static, (ii) passive counter weight based (iii) passive spring based, and (iv) active systems. (Reproduced from [1])

systems continually measure the interaction force between the patient and the BWS actuator and based on these measurements, a control system commands the actuator to move in such way that a constant amount of vertical force is felt by the patient, despite the vertical movements of the patient during walking. With a high enough control bandwidth, these systems can provide comfortable weight unloading to promote natural walking [59, 60].

Robot assisted devices available for gait rehabilitation feature some form of BWS system. HapticWalker [61] is an end-effector type device that utilizes a passive trunk suspension module to unload patients while walking. The BWS system keeps the absolute position of the human CoM constant by controlling the trajectory of the foot plates. MIT Skywalker [62] is another end-effector type device that features a loose chest harness to prevent patient falls and a passive spring based BWS system that uses a bicycle seat to unload patient weight. The position of the seat is pre-set by moving a linear actuator up or down using a remote control to allow for the desired level of weight support. The gait rehabilitation system comprising of POGO and PAM [63] uses the commercial Robomedica active BWS system that connects to the patient through an overhead harness and controls the tension of the overhead cable. Lokomat [1] utilizes Lokolift active dynamic BWS system. The patient is connected to an overhead harness that connects to a spring through pulleys. The unloading amount is measured by a force sensor and a motor adjusts the spring length to compensate for the desired level of patient weight. WalkTrainer [52] is an overground gait rehabilitation device with an active BWS system that consists of a controlled preloaded spring whose length is adjusted according to force sensor measurements. This system also connects to the patient through an overhead harness. KineAssist [64] gait and balance training system supports the body weight using an active custom designed harness that connects at the pelvis of the patient The unloading force is measured using load cells embedded in the harness and desired level of unloading is implemented using a force controller. NaTUre-gaits [65] is a hybrid device that implements end-effector (foot plate) type technology with a mobile base to deliver control of foot movements and an overground walking experience. It features active BWS at its pelvis module that measures the interaction forces using force sensors and actively generates the desired amount of weight unloading by creating a virtual spring along the pelvic trajectory. All of the active BWS systems presented above rely on stiff commercial load cells for closed loop force control and neither of them features an inertial compensation scheme. AlterG Anti Gravity Treadmill [66] functions by creating a difference in air pressure around the lower and upper body of the patient by inflating. The pressure inside a sealed bag around the lower body is adjusted to keep constant support as a percentage of the patient's body weight. This device does not feature inertia compensation. ZeroG [60] is an overground BWS system that moves on overhead rails and implements a force control strategy based on SEA. In particular, this system utilizes SEA to measure and actively control the tension in the rope that connects to the patient harness. This system does not implement an inertia compensation strategy, instead considers inertial forces due to the device as disturbances and relies on a PI force controller in an attempt to overcome the force deviations caused by these forces. ZeroG does not allow for lateral movements; thus, can cause balance problems during overground walking [67,68]. FLOAT [68] is a multi degree of freedom version of ZeroG that consists of two overhead rails connected to the patient harness at four points so that it can allow patients to move laterally in addition to the forward/backward movements of ZeroG. Similar to ZeroG, FLOAT does not compensate for inertia of the human body while moving.Compensation for inertial forces caused by vertical movement of patient body has been mostly ignored in BWS systems available for gait rehabilitation. However, considering that the amplitude of vertical CoM displacements is about 49.8 ± 10.3 mm at natural gait cycles [69], the inertial forces can account up to 25% of the gravitational loads.

In line with the requirements of effective body weight support, we have used Gravity-Assist which is a series elastic active body weight support system with inertia compensation. More details are discussed in the relevant chapter.

Chapter III

3 AssistOn-GAIT Overground Gait and Balance Trainer

The overall design, kinematics and implementation of the proposed overground gait and balance trainer is presented in the chapter.

3.1 Design

AssistOn-GAIT (Figure 3.3 ,3.4 and 1.3) is a robotic rehabilitation device designed to administer overground/treadmill based gait and balance rehabilitation training to adult patients with gait disorders. The device actively controls 7-DoF of the pelvis hip complex which include the forwards/backwards pelvic displacements and pelvic rotation in transverse plane, vertical pelvic displacement, pelvic tilt in coronal plane,rotations of the each hip in the sagittal plane and lateral pelvic displacement (Figure 1.2). Other hip movements including the hip abduction/adduction and hip lateral/medial rotation are passively allowed. Consisting of a pelvis/hip exoskeleton (EXO-with motion transmitter, human connection and a lateral pelvic actuation mechanism (LPAM)), an active body weight support (BWS) system and a mobile base, the device allows all pelvis/hip movements and can apply guidance forces along the seven crucial degrees of freedom. Overall kinematics of the device are shown in Figure 3.1 and 3.2. Pelvis-hip movements covered by the device are shown in Figure 4.8.

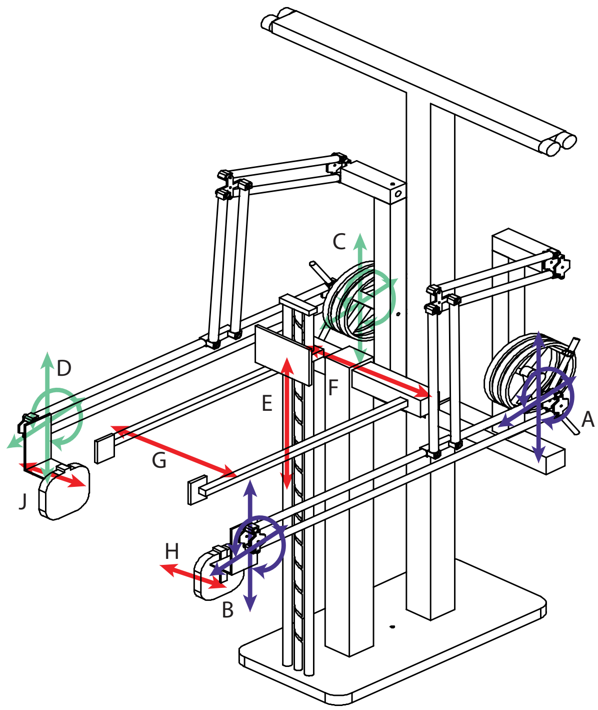

Figure 3.1: Kinematics of AssistOn-GAIT [treadmill version] (A-B) Left 3$\underline{\text{R}}$RP mechanism and its motion transmitted to human connection (C-D) Right 3$\underline{\text{R}}$RP and its motion transmitted to human connection (E) Movements of BWS system (F-G) Movement of LPAM (H-J) Passive lateral movement of human connection

3.2 Implementation

The pelvis-hip exoskeleton module (EXO) of AssistOn-GAIT consist of two planar parallel 3$\underline{\text{R}}$RP [1] mechanisms connected to patient through a custom motion transmitter and connection mechanism. The EXO module can

[1]P represents a prismatic joint, R represents a revolute joint

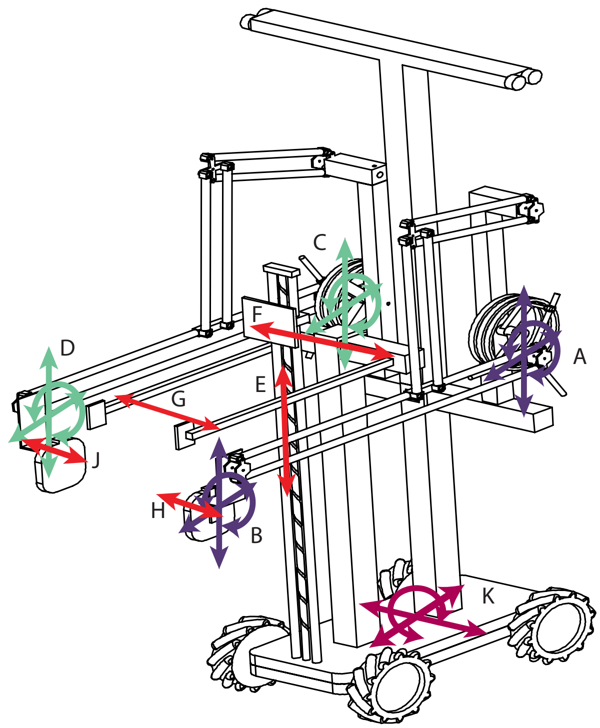

Figure 3.2: Kinematics of AssistOn-GAIT [mobile version] (A-B) Left 3RRP mechanism and its motion transmitted to human connection (C-D) Right 3RRP and its motion transmitted to human connection (E) Movements of BWS system (F-G) Movement of LPAM (H-J) Passive lateral movement of human connection (K) 3DoF movement of the mobile base

independently actuate 6 DoF of the pelvis-hip complex: the rotations of hip joints in the sagittal plane, the pelvic rotation in the transverse plane, the pelvic tilt in the coronal plane and the vertical pelvic displacement. Furthermore, the pelvis-hip exoskeleton module is passively backdriveable and self-aligning. Passively backdriveable design of the pelvis-hip exoskeleton module enables both passive movements of these DoF and their independent active control, while keeping the minimum impedance model interaction forces and

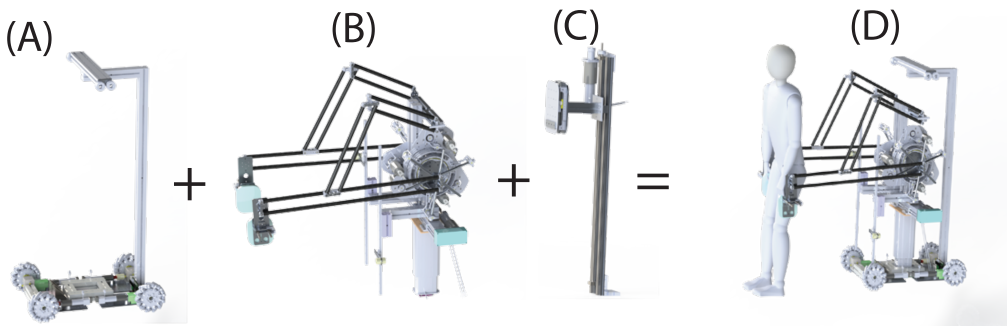

Figure 3.3: Modules of the Overground version of AssistOn-GAIT (A) Mobile base (B) Pelvis-hip exoskeleton (C) Body weight support system (D) Full system

torques low. Automatically adjusting the center of rotation of its joint axes, pelvis-hip exoskeleton module enables an ideal match between patients hip rotation axes and the device axes in the sagittal plane, and can do so while allowing for natural pelvic movements during walking. This feature not only guarantees ergonomy and comfort throughout the therapy, but also extends the usable range of motion for the hip joint. Moreover, the adjustability feature significantly shortens the setup time required to attach the patient to the exoskeleton.

Lateral movement of the pelvis is handled by the Lateral Pelvic Actuation Module (LPAM). The LPAM module is an admittance controlled 1 DoF system that actively controls the lateral pelvic movements. These modules are augmented by the BWS, which is also a 1 DoF admittance controlled system that is able to support a desired percentage of the patients' weight while ensuring proper balance and posture. Details of the BWS are mentioned in the relevant chapter.

Hence, overground gait and balance exercises, as well as pelvis-hip ex-

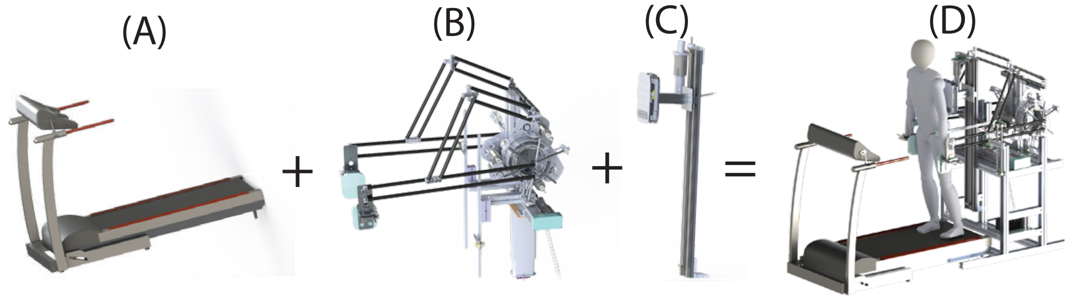

Figure 3.4: Modules of the Treadmill version of AssistOn-GAIT (A) Treadmill (B) Pelvis-hip exoskeleton (C) Body weight support system (D) Full system

ercises intended to correct compensatory movements arising from abnormal gait patterns can be delivered with AssistOn-GAIT extending the type of therapies that can be administered using lower extremity exoskeletons. The workspace of each module is shown in Table 3.1, while the force rendering capability is given in Table 4.2. The range of pelvis-hip movements covered by the device is shown in Table 4.1.

Table 3.1: Workspaces

Module	Axis	Workspace
EXO	along the sagittal axis	450 mm
	along the longitudinal axis	450 mm
	about the horizontal axis	300°
LPAM	along the horizontal axis	300 mm
BWS	along the longitudinal axis	1000 mm

3.3 Control

As explained above, AssistOn-GAIT features a modular design where the EXO, BWS and LPAM work in coordination with each other. The EXO is impedance controlled, while the BWS and LPAM are admittance controlled. Real time communication is implemented using EtherCAT, which has been selected due to its scalability, ease of programming and robustness. The processing is handled by a host computer running Simulink Real Time operating system. Each motor is controlled by fast digital controllers that operate on 10 kHz. Control strategy for individual modules of AssistOn-GAIT are given in the subsequent sections.

In the rest of the report, we present the EXO, LPAM, BWS and mobile base modules. It is important to mention here that the proposed device can be utilized for treadmill based as well as for overground training.

Chapter IV

4 Self Aligning Pelvis-Hip Exoskeleton Module

This chapter discusses the design and control of EXO module of AssistOn-GAIT . The self aligning exoskeleton enables controlling 7 DoF of the pelvis-hip. The DoF are shown in Figure 4.8. It consists of these modules:

- Dual 3$\underline{\text{R}}$RP mechanisms
- Motion transmitter mechanism
- Human pelvic connection
- In plane mechanism
- Gravity compensation
- Lateral pelvic actuation mechanism (LPAM)

The self aligning pelvis hip exoskeleton is explained in the paragraphs below.

4.1 Dual 3$\underline{\text{R}}$RP Mechanisms

4.1.1 Design

The 3$\underline{\text{R}}$RP mechanisms of the exoskeleton actively control 6 DoF of the pelvis-hip complex in the sagittal plane (the remaining one, lateral displace-

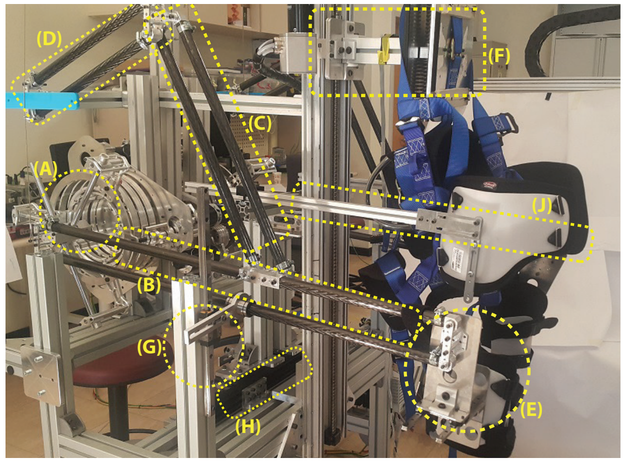

Figure 4.1: Modules of the Exoskeleton [(A) 3$\underline{R}$RP mechanism (B)-(D) Motion transmitter (E) Pelvis human connection (G) In plane mechanism], BWS [(F)], LPAM [(H),(J)]

ment, is controlled by the LPAM). These pelvis-hip movements are crucial for gait rehabilitation, however, are not safety critical in prevention of falls. It is also required that these DoF be impedance controlled with the possibility of generating haptic guidance that can guide patient movements for correction of compensatory movements. In view of this an impedance controlled mechanism is preferred which features a low friction and inertia. For these design requirements two 3$\underline{R}$RP parallel mechanisms have been selected for the exoskeleton. The mechanisms have been presented in [70–72] as parts of self-aligning knee and shoulder exoskeletons. A modified version of the parallel 3$\underline{R}$RP mechanism posessing a larger workspace and torque force rendering capabilities have been designed for the EXO. The mechanisms are designed to have a workspace and force rendering that covers the full range of motion required for healthy and impaired patients.

4.1.2 Implementation

The 3$\underline{\mathrm{R}}$RP Mechanisms (Figure 4.2) are symmetric parallel mechanisms that possess a large, circular, singularity free workspace. The grounded revolute joints of each mechanism are actuated and enable two translational and one rotational DoF in the sagittal plane. All these DoFs can be controlled independently or in a coordinated manner. All DoF are equipped with low friction dual layer power transmission and feature high passive backdriveability. Thanks to their parallel kinematics, the 3$\underline{\mathrm{R}}$RP mechanisms not only feature high bandwidth and stiffness, but also serve as a mechanical summer during end-effector rotations. Therefore, relatively small actuators can be used to impose large torques and forces at the end-effector of mechanism.

The 3$\underline{\mathrm{R}}$RP mechanism design is realized using three concentric custom made rings that move on slim bearings. The rings are actuated using three 48 V, 250 W DC motors (fitted with 2000 counts per turn incremental encoders) via a dual layer capstan transmission with an overall reduction ratio of 1:30. Each ring is connected to custom made aluminum brackets that hold cylindrical aluminum rods. In the center, the rods are connected to collocated bearings that form the end effector. Overall, the system is capable of generating 135 N force (3000 N instantaneous) along its translational DoFs and 72 Nm (1440 Nm instantaneous) torque along its rotational DoF. The end effector can cover 450 mm along its translational DoFs and 300° about the horizontal axis . The specifications are listed in Table 3.1 and Table 4.2. Hence, the kinematics of the pelvis-hip exoskeleton enable independent actuation of each hip flexion/extension, the pelvic tilt, the pelvic rotation, the vertical and horizontal translations of the pelvis. End effectors of each 3$\underline{\mathrm{R}}$RP mechanism are connected to a motion transmitter.

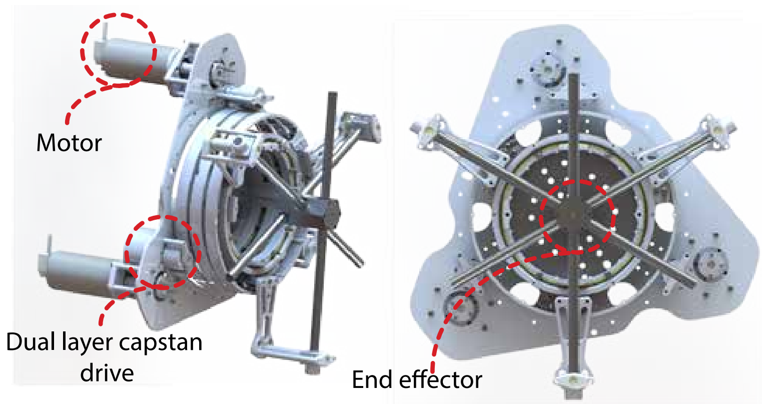

Figure 4.2: CAD model of the self-aligning 3$\underline{R}$RP mechanism

Table 4.2: Force rendering capability of AssistOn-GAIT

Module	Axis	Force/Torque
EXO	along the sagittal axis	135 N (3000 N instantaneous)
	along the longitudinal axis	135 N (3000 N instantaneous)
	about the horizontal axis	72 Nm (1440 Nm instantaneous)
LPAM	along the horizontal axis	240 N
BWS	along the longitudinal axis	1000 N

4.1.3 Control

The EXO operates under an open loop impedance control strategy as shown in Figure 4.3. This allows generation of haptic trajectories along each of the three DoFs, that can guide the pelvis-hip for rehabilitation assistance purposes, the capability of which has been demonstrated in [70–72]. Furthermore, independent impedance control of each degree of freedom allows the generation of virtual fixtures along a desired pelvis-hip trajectory. Once the desired pelvis-hip trajectory has been determined based on patient specific therapy requirements, it is modeled as a Non-uniform rational basis spline (NURBS) curve. Modeling as a NURBS curve allows generation of complex shaped trajectories that can truly represent how the pelvis-hip moves in the sagittal plane, and utilize closest point tracking to determine the relative position of the pelvis-hip to the nearest point on the curve. To represent virtual fixtures, two offset NURBS curves are generated, the inner one representing the inner fixture and the outer representing the outer fixture. As the pelvis-hip moves, real time calculations are made to determine if it is between fixtures or contacting any of the fixtures. In case of contact, opposing haptic forces are generated to correct the pelvic trajectory. This is a very important feature as it does not dictate a time based trajectory to the connected human but the human is free to set their own pace of walking. A demonstration of this control is shown in the Experimentation section of the paper.

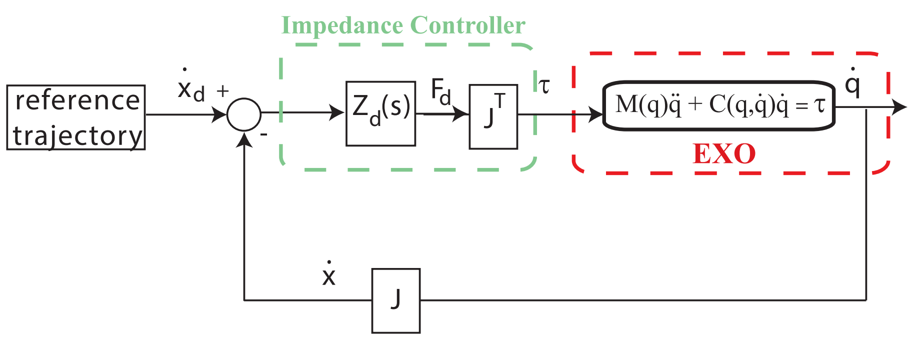

Figure 4.3: Open loop impedance control for the EXO (J is the Jacobian, Z_d is the desired impedance, q is the joint angle vector, x is the end effector position vector, F_d is the desired end effector force vector, τ is the desired motor torque vector, M is the EXO inertia matrix, C is the EXO Coriolis matrix)

4.2 Motion Transmission and Human Connections

4.2.1 Design

While walking humans swing their arms in repetitive patterns which vary for each individual. The utilization of arm swinging is more important when the body gets imbalanced (for example tripping while walking) and helps with regaining the lost balance [77]. As normal walking is not always performed on straight smooth surfaces, there may be many instances when the body tends to lose balance. While it may be natural to recover from these situations by healthy people, impaired patients would find it challenging. Research also indicates that arm swing helps reduce the metabolic cost of walking [77–79]. Therefore, it is considered that arm swinging should be allowed for effectiveness of gait rehabilitation. Many of the recent RAGT devices however, restrict arm swinging due to their design constraints. This includes the Lokomat [80], LOPES I [?], NaTUre-gaits [?], Motorika Reoambulator [43], ALEX [44],RGR [?] and Walk Trainer [51, 52]. Other devices that allow arms swing are the LOPES-II [39] and PAM [40]. In line with the importance of Arm Swinging, AssistOn-GAIT features a human connection module that enables the connected patients to swing their arms freely.

4.2.2 Implementation

The EXO 3$\underline{\text{R}}$RP mechanisms actuate 6 DoF of the left and right pelvis-hip. Connecting the human directly to these mechanisms would impede the free arm swing movement. To cater for this, the complex 3$\underline{\text{R}}$RP mechanisms have been moved back and their 3 DoF (x, y, θ) motion is carried to the human through the Motion Transmitter module. The module starts from the respective $3\underline{R}RP$ end effectors and ends at the Human Connection. Purpose of this module is to transmit two translation and one rotational DoFs from the $3\underline{R}RP$ to the human connection. Due to the $3\underline{R}RP$ having 3 DoF, a total of three parallelograms have been used. Parallelogram mechanism is ideal for preserving rotational motion, however, translational motions also need to be preserved. Therefore, one of the parallelograms needs to be grounded, such that the parallelogram connected to the $3\underline{R}RP$ end effector is always parallel to the ground. It is constructed using light weight carbon fiber rods that keep the weight and inertia low while maintaining enough strength to transfer forces and torques. The kinematics are shown in Figure 4.4. Out of plane movements of the motion transmitter are restricted using a passive XY slider mechanism. The Human Connection modules are installed at the end of Motion Transmitters.

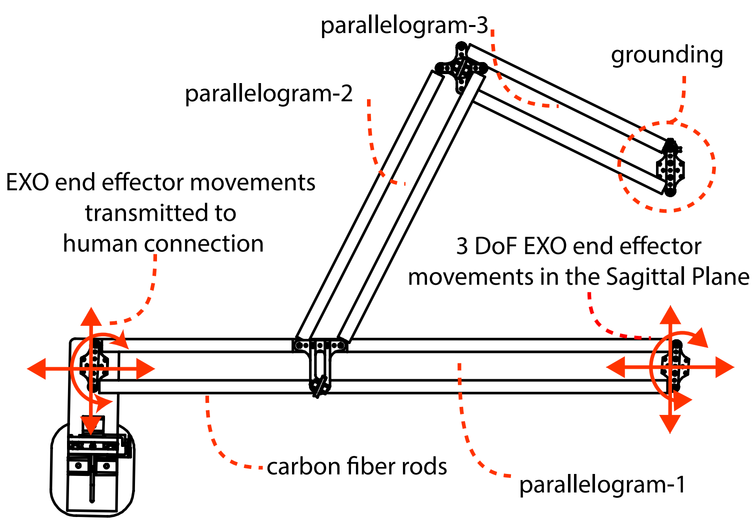

Figure 4.4: Motion transmitter kinematics

At the end of each Motion Transmitter Module, is the *Human Connection Module*. The purpose of this module (Figure 4.5) is the connect to the patient at the left and right hips, and transfer movements from the motion transmitter in the sagittal plane. This module allows the lateral pelvic movement passively and enables the human to swing both arms freely.

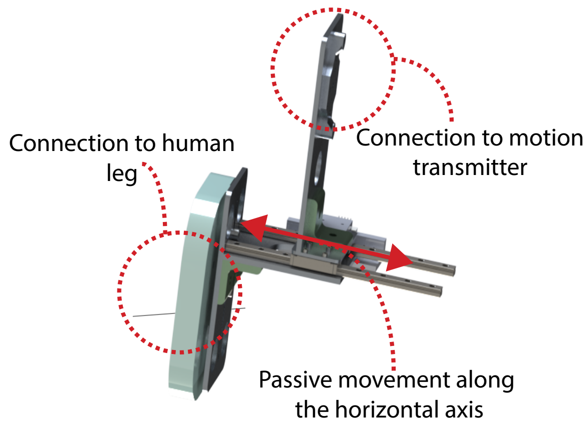

Figure 4.5: Human Pelvic Connection

Actuation of the EXO is performed by the left and right 3$\underline{R}$RP mechanisms which are planar mechanisms operating in the sagittal plane, and must not be subjected to out of plane forces, in order to work optimally. The transmitter mechanism connected to the end effectors of 3$\underline{R}$RP mechanisms tends to induce a lateral force and a large torque in the transverse plane. A custom XY slider mechanism has been designed to ensure that the motion transmitter moves in the sagittal plane only. The system is shown in Figure 4.6.

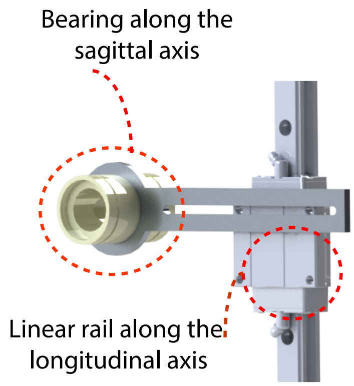

Figure 4.6: Restricting out of plane movements

4.3 Passive Gravity Compensation

4.3.1 Design

Weight of the 3$\underline{R}$RP mechanism end effector, motion transmitter and human connection tend to pull the EXO downwards. In order to keep the mechanism in equilibrium, a gravity compensation system is mandatory. Two major methods of gravity compensation are *active compensation* where actuators are used to actively balance the effects of gravity, and *passive compensation* where passive elements such as springs are utilized to counter effects of gravity. Keeping in view the requirement of passive backdriveability for the EXO, a passive gravity compensation system has been designed and implemented.

4.3.2 Implementation

We have placed special emphasis on reducing complexity in the system, and the gravity compensation has also been designed accordingly (Figure 4.7).The desired compensation is generated by two constant force springs installed in parallel to each other. A single cable runs down from the springs to the EXO end effector. The springs provide an upwards force along the longitudinal axis which is equal to the combined weight of the 3$\underline{\mathrm{R}}$RP end effector, motion transmitter and pelvis connection. The springs are installed very high above the EXO end effector so that the contribution of the compensation force in the horizontal direction (along the sagittal axis) can be kept to a minimum. It has been calculated that when the 3$\underline{\mathrm{R}}$RP end effector is at a distance of 100 mm from the origin along the sagittal axis, the component of the compensation force that acts along this axis is only 0.07 times the total compensation force.

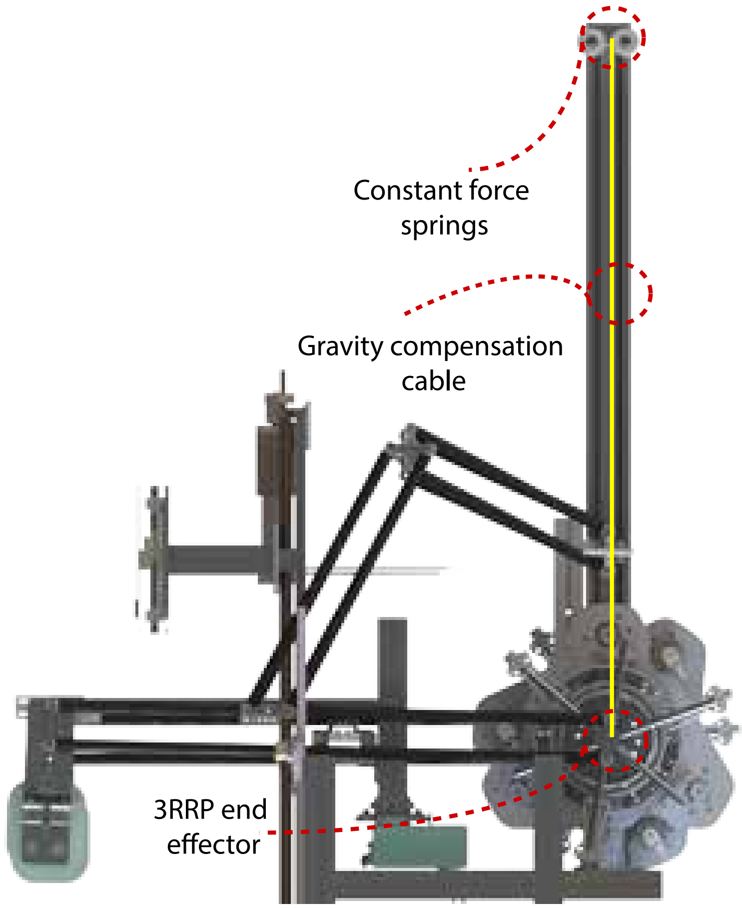

Figure 4.7: Gravity compensation system

4.4 Lateral Pelvic Actuation Mechanism

4.4.1 Design

As discussed in the previous sections, the lateral pelvic displacement being one of the determinants of gait is extremely important for effective gait rehabilitation. Walking when the lateral pelvic displacement is restricted, results in reduced range of motion for the lower limb [53]. In conventional manual gait rehabilitation therapy or in BWSTT, a human therapist is dedicated to moving the pelvis in the lateral direction. Many RAGT devices feature active lateral pelvic actuation. Among these are the Lokomat which has the optional FreeD module that allows for lateral translation of the pelvis [80], while the interaction force with the connected patient is sensed using force sensors. The Kineassist [50] uses a parallelogram mechanism to allow for lateral translations passively. The WalkTrainer [51, 52] can actively monitor and control interaction forces in the lateral direction by using force sensors. PAM [40] utilizes pneumatic actuators for active control of lateral pelvic displacements. An overground trainer presented by [53] also utilizes force sensors to measure and control interaction forces.

For lateral motion of the pelvis, high interaction forces are required. Furthermore, when performing balance training the patient may tend to lean towards one side, and safety demands using a mechanism with high force output. In line with this requirement, an actively backdriveable admittance type mechanism is favored. For sensing of interaction force, when stiff force sensors are used, the robustness of force control is compromised and the output impedance is very high. These problems can be overcome by using series elastic actuation, at the cost of a lower bandwidth. This technique has been discussed in detail by [15]. This makes series elastic actuation most suitable for robot human interaction. Due to this reason, we propose that a series elastic actuated lateral pelvic guidance is the most suitable and can become a true replacement of a human therapists (as human hands are not stiff but compliant). The amount of lateral interaction force between human therapists and hemiplegic patients has been determined to be up to 20 N [81]. However, patients with different types of impairments may require more force in the lateral direction, which has not been characterized in literature. Keeping this in view AssistOn-GAIT has been designed to provide up to 240 N nominal force in the lateral direction.

4.4.2 Implementation

The LPAM handles the lateral pelvic movement as shown in Figure 4.10. It consists of a one degree of freedom series elastic actuator which connects to the human at the hip. The amount of lateral force exerted on the pelvis can be measured and actively controlled. Being series elastic actuated, the module utilizes low stiffness springs and features a relatively low mechanical output impedance.

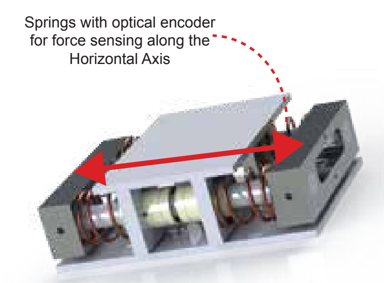

Figure 4.9: Lateral compliant element for series elastic control

The lateral workspace is $\pm 150mm$, and can apply continuous force of up to 240 N. The series elastic unit is shown in Figure 4.9.

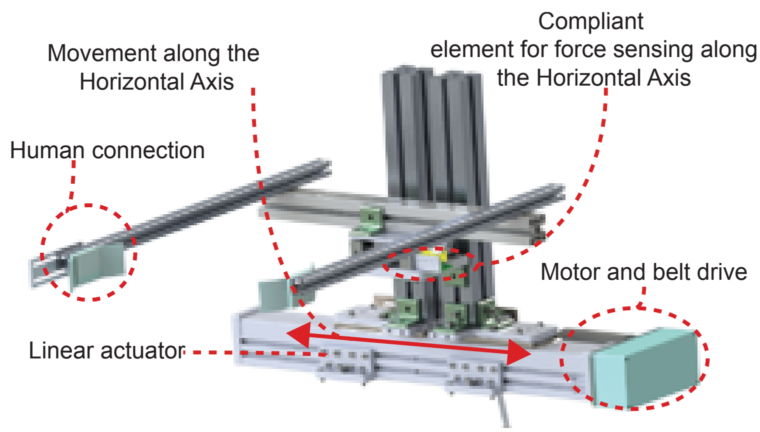

Figure 4.10: Lateral Pelvic Actuation Mechanism (LPAM)

4.4.3 Control

A real-time cascaded admittance controller is implemented for SEA, as shown in Figure 4.11. This way, the LPAM can be programmed to render desired admittance values.

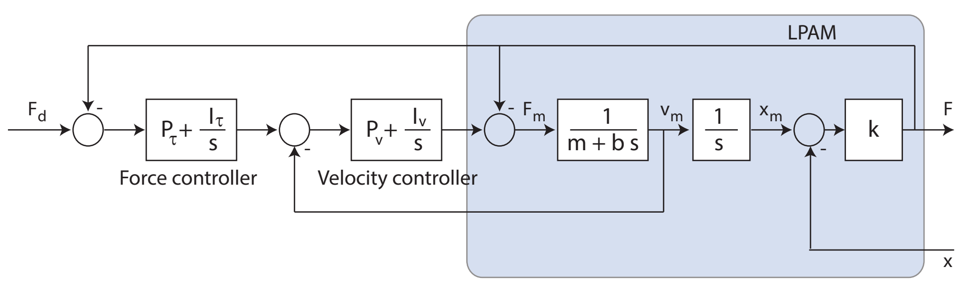

Figure 4.11: Cascaded admittance controller of the LPAM

In the figure P_τ and I_τ represent the gains of the outer PI force controller, P_v and I_v represent the gain of the inner PI velocity controller, m and b represents the inertia and damping of the device, k is the stiffness of the compliant element. The signal F_d represents the desired force, F_m is the motor force, F is the actuator output force, x is the displacement caused by human. The cascaded controller consists of an inner robust velocity control loop and an outer force control loop. The inner loop of the control structure deals with imperfections, such as friction and stiction of the linear actuator, rendering the system into an ideal velocity source. The velocity controller is implemented in hardware on the motor driver with fast control rate of 20 kHz. The outer force control loop is implemented at 1 kHz for high fidelity force control. This loop imposes the desired level of support forces as determined by the therapist.

Table 4.1: Pelvis-hip movements covered by AssistOn-GAIT(Supported Hip widths 270-420 mm, Supported Hip heights 653-1202 mm, covers ranges published by NASA as part of Anthropometry and Biomechanics Data [3])

Degree of freedom	Status	Delivered By	Achievable Range	Human Range
Vertical pelvic displacement	Active	Exo/BWS	450 mm	$3.7 \pm 0.8\%$ of leg length [73]
Lateral pelvic displacement	Active	LPAM	± 150 mm	$\pm 40.8 mm$ [74]
Fwd/back pelvic displacement	Active	Exo	Unlimited	Unlimited
Pelvic tilt in coronal plane	Active	Exo	$\pm 12°$	$9.4° \pm 3.5°$ [73]
Pelvic rotation in transverse plane	Active	Exo	$\pm 10°$	$\pm 6°$ [75]
Hip flexion/extension	Active	Exo	300°/300°	30°/10° [76]
Hip abduction/adduction	Passive	Human connection	–	–
Hip lateral/medial rotation	Passive	Human connection	–	–

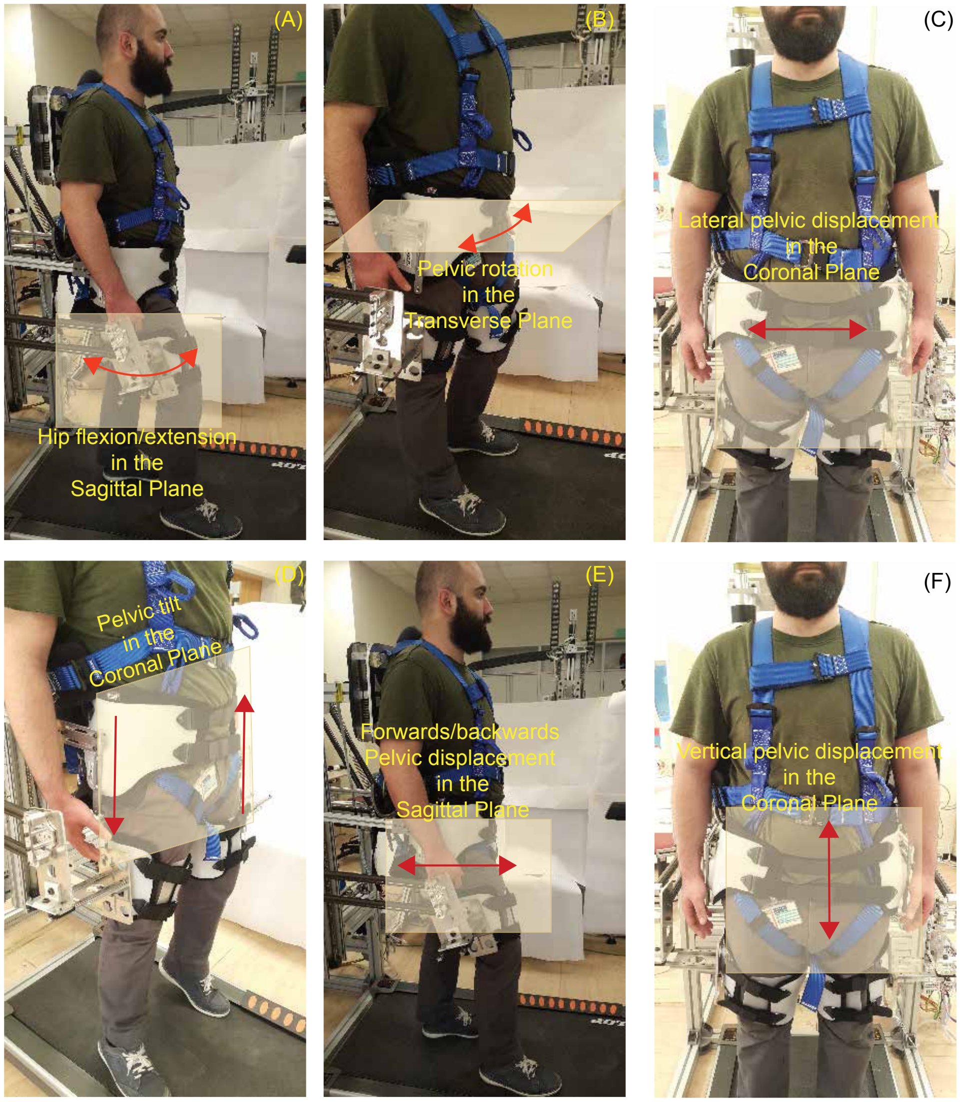

Figure 4.8: Kinematics of AssistOn-GAIT (A) Left and right hip flexion-extension (B) Pelvic rotation (C) Lateral pelvic displacement (D) Pelvic tilt (E) Forwards-backwards pelvic displacement (F) Vertical pelvic displacement

Chapter V

5 Holonomic Mobile Base

In the mobile version of AssistOn-GAIT (Figure 3.3), the holonomic mobile base carries all the hardware including the exoskeleton and body weight support system. The mobile base must be able to move with the connected patient and along 3 DoF on the ground. Furthermore, the inertia of the mobile base should not be felt by the connected patient. To ensure this we explore two control techniques for the mobile base namely *series elastic actuation* and *workspace centering control*. In this chapter, and introduction to the mobile base is presented followed by the design of the mobile base for both control settings, implementation, characterization and experimentation.

5.1 Holonomic Base with Mecanum Wheels

The mobile platform possesses three DoF, two DoF in translations and one DoF in rotation, to sustain all possible overground movements. The mobile robot is chosen as of holonomic type, so that it is omnidirectional. Redundant actuation with four Mecanum wheels – omni-directional wheels with 45° angled rollers – is preferred, since it allows for lower power motors be utilized to achieve high forces/torques outputs at the task space of the robot together with smooth holonomic movement with enhanced traction.

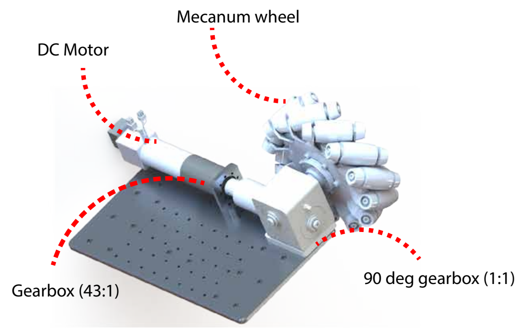

Figure 5.1: Module of the Holonomic mobile base

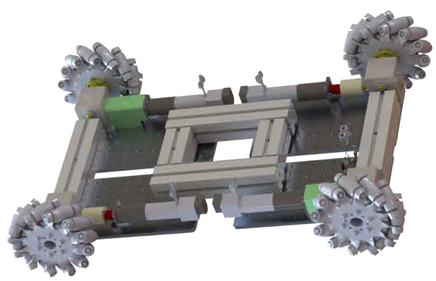
Figure 5.2: Holonomic mobile base

The mobile base is designed to carry the weight of exoskeleton, body weight support and associated hardware while being quick and precise in its movements to allow for effective control. The base has been designed as four modules connected to each other using aluminum profiles. This modular construction allows for changing the dimensions of the base quickly and effortlessly to suit the hardware that is placed on top of it. Design of the module is shown in 5.1. Each module consists of:

- DC Motor
- Gearbox (43:1)
- 90 degree gearbox (1:1)

- Mecanum wheel

Each module has a brushless DC motors attached to Mecanum wheels through gearboxes. Each motor with gearbox can deliver a torque of 7 Nm and reach a maximum of 213 RPM. Control of motors is performed in hardware using hardware based PID control operating at 10 kHz. When the four modules are combined together all four motors can deliver a combined torque of 28 Nm, which is able to move the mobile base with associated hardware and 50 kg human weight at a maximum speed of 0.4 m/s, which is sufficient for gait rehabilitation purposes. A prototype of the mobile base of AssistOn-GAIT is shown in Figure 5.2. The velocity control bandwidth of the holonomic base is experimentally characterized as 6 Hz as shown in Figure 5.3, which is sufficiently higher than the desired force control bandwidth of the mobile base dictated by the speed of human movements.

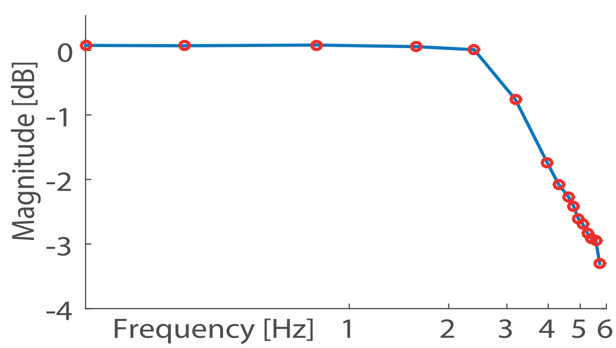

Figure 5.3: The closed-loop position control bandwidth of the holonomic mobile base is experimentally characterized as 6 Hz.

5.2 Series Elastic Actuation of Holonomic Mobile Base

Although the actuators and power transmission integrated to the mobile platform are fairly backdriveable, the use of Mecanum wheels impose passive non-backdriveability to the overall system. To be able to achieve active backdriveability, the proposed mobile device is turned into a multi-DoF series elastic actuator with the integration of a multi-DoF compliant element. This approach has been introduced for upper-extremity robotic rehabilitation therapies in [2, 82].

Series Elastic Actuators (SEAs) are formed by intentionally introducing compliance between the actuator and the load. This compliance presents a number of advantages including low output impedance and better shock tolerance. From a control systems design perspective, SEAs allow higher controller gains (as compared to using force sensors) and thus more accurate/robust force control within the control bandwidth. Measurement of output force can be easily done by sensing the deflections of the compliant element, thus eliminating the need for expensive and noisy force sensors. The main disadvantage of SEA is the reduction in the system bandwidth. However, since the movements during rehabilitation are slow, SEAs are ideal for rehabilitation devices that operate in contact with slow moving patients.

To achieve the benefits of SEA for AssistOn-GAIT, a multi-DoF compliant element has been introduced between the patient and the mobile base, such that the mobile platform behaves as a three DoF SEA. This enables accurate measurement and control of the interaction forces between the patient and the mobile base as an essential aspect of effective rehabilitation therapies. Figure 5.4 presents the multi-DoF compliant element implemented as a monolithic 3PaRR[2] planar parallel mechanism. For small deflections, the pseudo-rigid body model of the compliant mechanism can be utilized with high accuracy and the deflections of the grounded Pa joints measured via high resolution optical encoders can be mapped to forces/torque applied at the moving platform of the compliant mechanism. The compliant mechanism is supported by a 3 DoF guide that counteracts non-planar forces/moments and mechanically limits excessive deflections of the compliant element.

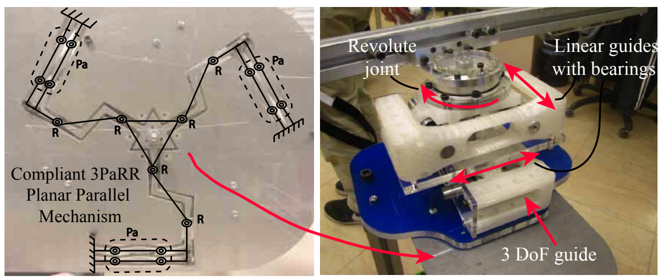

Figure 5.4: The multi-DoF compliant element is implemented as a monolithic 3PaRR planar parallel mechanism [2], while a 3 DoF guide is utilized to counteracts non-planar forces/moments and to limit excessive deflections of the compliant element.

To ensure backdriveability of AssistOn-GAIT under the action of forces applied by the patient at the end-effector of the compliant element, a real-time controller is implemented. The block diagram of the closed loop SEA system is presented in Figure 5.5, where physical variables are marked with thicker lines. In this cascaded structure, there is an inner velocity control loop and an outer force/admittance control loop. The inner loop of the control structure deals with imperfections such as friction and stiction, rendering the system into an ideal velocity source. For the outer loop, an admittance

[2]Pa represents a parellelogram linkage, R represents a revolute joint

controller is implemented. In particular, in Figure 5.5, F_{human} represents the forces applied by the patient, τ is the torques applied by the actuators of the holonomic mobile platform, while $\dot{q}_h$ denotes the velocities of these actuators. Symbols K_p and K_i denote the PI gains of the inner-loop velocity controller of the mobile platform, whereas M_a and b_a represent the parameters of the desired admittance. The symbol K_T is the task space stiffness, while $J_{T_{comp}}$ represents the overall Jacobian of the compliant 3PaRR mechanism. The symbol Δx_T represents the task space deflections, while Δq_m denotes the joint space deflections. Finally, J_h is the Jacobian of the holonomic platform, while $J_h^\sharp$ represents its pseudo-inverse.

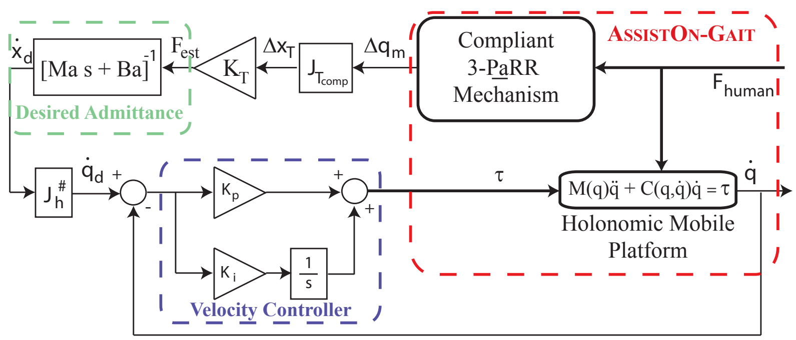

Figure 5.5: Cascaded admittance controller of the series elastic mobile base of AssistOn-GAIT

Active backdriveability of AssistOn-GAIT is tested through several feasibility experiments with healthy volunteers, where a virtual tunnel is introduced between ∓ 150 mm at x-direction, defining forbidden regions in the task space. Figure 5.6 presents sample results from these experiments, where the path followed by AssistOn-GAIT is depicted along with the direction of forces applied by the volunteer. From this figure, it can be observed that the movement of the mobile base of AssistOn-GAIT closely follows the direction of forces applied to it. However, whenever the patient reaches the boundary of the virtual tunnel, the controller gently guides the device inside the tunnel by rendering opposing forces to satisfy virtual constraints. The stiffness of the virtual constraints as well as the apparent impedance of the device within the control bandwidth can be adjusted through the controller gains.

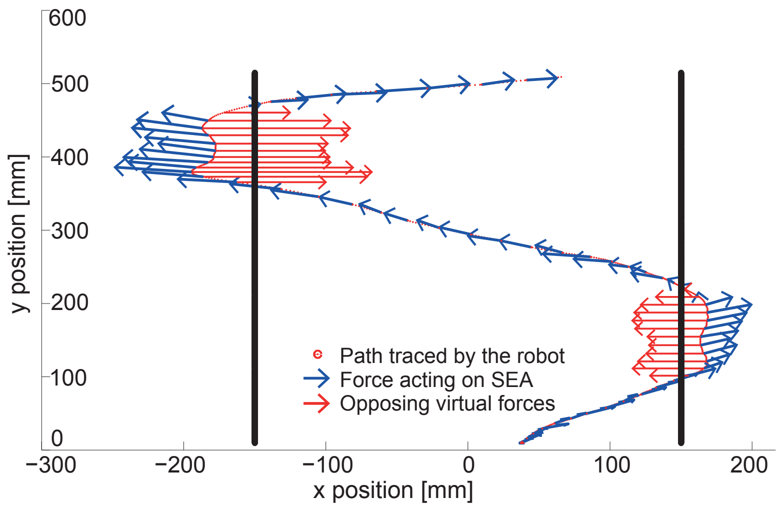

Figure 5.6: Virtual fixtures are implemented at $x = \pm 150$ mm. The smooth movements of the device under user applied forces and guidance forces rendered due to the virtual fixtures are presented.

5.2.1 Drawbacks of Series Elastic Control of Mobile Base

Series elastic control of the mobile base works well however as explained above, however, it has its limitations. Force control bandwidth limits the force rendering performance of series elastic actuation. Furthermore, even small movements by the human cause the mobile base to move. To overcome these difficulties we propose Workspace Centering Motion Control. To implement this control, Redundant active DoF is added to Exoskeleton along lateral pelvic direction. The overall system becomes a kinematically redundant system (exoskeleton can cover all DoF of mobile base) which is important for transparent operation. The Exoskeleton has required DoF but its workspace is limited, and the mobile base has the same DoF but has a larger workspace. This enables Dual layer (micro-macro) actuation. With a fast enough control of the mobile base, the human will only feel presence of exoskeleton.

5.3 Workspace Centering Control of Mobile Base

In this section, we present the kinematically redundant design and work space centering control of AssistOn-GAIT, an overground gait and balance trainer designed to deliver pelvis-hip exercises to correct compensatory movements arising from abnormal gait patterns. AssistOn-GAIT consists of a force / impedance controlled pelvis-hip exoskeleton module that can assist pelvic movements of patients, attached to a motion controlled holonomic mobile platform that allows patients to walk naturally on flat surfaces, start/stop motion, vary their speed, sidestep to maintain balance and turn to change walking direction.

5.3.1 Background and Related Work

Robots designed for gait rehabilitation can be loosely categorized into three groups: end-effector type, exoskeleton type with treadmill, and overground trainers with mobile bases. In this chapter we focus on overground trainers with regards to workspace centering and inetria compensation.

Overground gait trainers possess mobile bases that follow/assist patients' walking movements on flat surfaces, allowing for gait practice under functional contexts and in combination with balance. These devices are advantageous since the patients move under their own control, while experiencing all the sensory inputs associated with walking. Patients can start/stop motion, vary their walking speed as desired, sidestep to maintain balance and turn to change walking direction. KineAssist [49,50] and Walk Trainer [51,52] are commercial examples of such overground gait trainers. Both devices consist of a mobile base connected to a pelvis orthosis and can provide partial body weight support. In particular, KineAssist is equipped with an admittance controlled Cobot base that compensates for the robot inertia based on forces measured at its custom designed torso-pelvis harness that allows for passive movements of the pelvis. KineAssist complies with natural user movements during walking but cannot assist pelvis/hip movements to help improve the quality of gait. Another overground trainer that relies on the interaction force measurements between the patient and the mobile base is proposed in [53], where unlike KineAssist, a holonomic mobile base is controlled according to these forces.

Walk Trainer features active pelvis and leg exoskeletons attached to a differential-drive mobile base [52]. Thanks to a parallel mechanism actuating all six DoF at the pelvis, the system has the ability to actively assist all the pelvic movements. Walk Trainer detects patient intentions regarding walking speed and heading utilizing two potentiometers and can control its differential-drive mobile base to follow straight and curved paths.

NaTUre-gaits [54] is a similar device, consisting of pelvis and leg modules connected to a differential-drive mobile base. This system employs dual three DoF actuated Cartesian planar robots at each side of the hip to assist pelvic movements. These Cartesian planar robots are also used to measure local pelvic motions and these measurements are mapped to walking speed and heading angle to control the differential-drive mobile base of NaTUre-gaits to follow straight and curved paths.

Both WalkTrainer and NaTUre-gaits have relatively complex mechanical designs and possess passively non-backdriveable power transmission that necessitates use of force sensors and active control algorithms to ensure synchronization between the mobile base, pelvis movements and leg rotations to achieve a natural gait for the patient. Furthermore, featuring differential drive mobile bases, both devices are non-holonomic and cannot allow for lateral movements, such as sidestepping.

In this chapter, firstly, we add a redundant active DoF to the exoskeleton module to enable lateral pelvic displacements without the need for the movements of the mobile base. This addition ensures that AssistOn-GAIT features dual layer (also called micro-macro) actuation [83, 84], since the exoskeleton module can now span all the DoF covered by its holonomic mobile platform. The redundancy is critical for transparency of AssistOn-GAIT, since this design decision makes sure that the reflected inertia of exoskeleton is independent of the inertia of the heavy mobile base that carries the weight of the patient and the exoskeleton [85]. Secondly, we implement a workspace centering motion controller [86–91] for the mobile platform based on pelvis poses measured by the exoskeleton module, such that the workspace limits of the exoskeleton module cannot be reached during overground training. This controller not only provides AssistOn-GAIT with a virtually unlimited workspace, but also decouples the dynamics of the mobile platform from the exoskeleton dynamics. Consequently, the force rendering performance and output impedance of AssistOn-GAIT is dictated only by the design of its exoskeleton module.

5.3.2 Kinematic Redundancy and Dynamic Decoupling

When the exoskeleton module is attached to the mobile base, the overall system possesses 3 redundant active DoF. In particular, the exoskeleton module already covers all the of motions of the holonomic mobile platform, resulting in the redundancy. This redundancy is a design choice and is critical for ensuring transparent force control of AssistOn-GAIT. Thanks to this redundancy, AssistOn-GAIT can be decomposed into two subsystems, referred to as mini and macro structures. The mini structure is defined as the smallest distal set of DoF that can completely span the operational space, while the macro structure connects the micro structure to the ground. For AssistOn-GAIT, the exoskeleton serves as the micro structure and the mobile base constitutes the macro one. Since, the reflected inertia of a kinematically redundant robot is upper bounded by the inertial properties of the micro structure [85], AssistOn-GAIT inherits the inertial properties of its exoskeleton module and is independent of the inertia of the mobile base. Given the high inertia of the mobile base, this inertial decoupling is crucial for archiving good force control performance within a large force control bandwidth.

Furthermore, it has been shown that if the friction/damping forces between the mobile base and the exoskeleton can be kept low, then the dynamics of the exoskeleton can be completely decoupled from the dynamics of the mobile base, as long as the workspace limits are not reached [86,90]. The friction/damping forces can be kept low by active compensation of exoskeleton dynamics under closed loop force/impedance control architectures or through mechanical design, by minimizing frictional losses at the power transmission elements. In our current implementation, the actuation of the hip-pelvis exoskeleton is based on direct drive actuation with capstan transmission and the overall system features low friction forces resulting in a excellent passively backdriveability [72].

5.3.3 Workspace Centering Control

In many situations, it is desirable to increase the workspace of an existing device to achieve better and more realistic interaction. Interactions of this type has been introduced in [83] and further explored in [84], where a high performance (micro) robot has been mounted on a larger (macro) robot in order obtain enhanced functionality. This concept has also been applied to haptic interaction [86–93] to enhance the workspace of small haptic devices. Furthermore, transparency characteristics of high fidelity force-feedback devices attached to mobile bases have been investigated in [86,90]. It has been shown that the force rendering capabilities of the overall device will be equivalent to that of the high fidelity force-feedback device, if the mobile base can keep the user always within the workspace limits of the force-feedback device and the friction/damping losses between the mobile base and the force-feedback device device can be kept low. It has also been discussed that the best performance from the coupled system can be achieved when the high performance (micro) robot is kept close to its ideal operating point through fast enough movements of the macro structure.

AssistOn-GAIT possesses micro-macro dual layer actuation, where an exoskeleton with high force control performance is mounted on a larger and stronger robot in order obtain enhanced functionality. Figure 5.7 presents the normalized manipulability contours of the 3$\underline{\text{R}}$RP mechanism of the exoskeleton module over its workspace. It can be observed from this figure that the highest manipulability is achieved when the end-effector of the mechanism is at the center of its workspace and manipulability decreases towards the boundaries of the workspace.

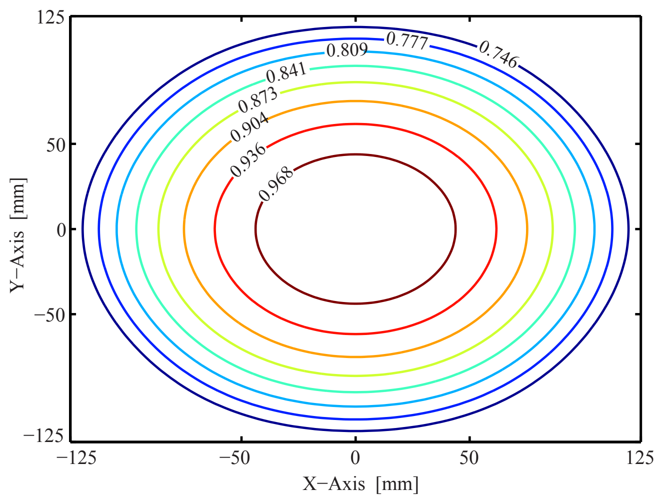

Figure 5.7: Manipulability of the 3<u>R</u>RP mechanism of the exoskeleton module

We implement a workspace centering motion controller [86–91] for the mobile platform of AssistOn-GAIT based on pelvis poses measured by the exoskeleton module, such that the workspace limits of the exoskeleton module cannot be reached during gait training. This controller not only provides AssistOn-GAIT with a virtually unlimited workspace, but having 6 Hz motion control bandwidth, this controller is fast enough to decouple the dynamics of the mobile platform from the exoskeleton dynamics during rehabilitation therapies. Consequently, since the workspace limits are never reached during gait and/or balance therapies, the force rendering performance and output impedance of AssistOn-GAIT is dictated only by the design of its exoskeleton module.

Note that the exoskeleton module can span all the DoF covered by the holonomic mobile platform and possesses enough workspace for at least one full stride. The goal of the mobile platform is to carry the weight of the patient and devices with minimal intervention. Furthermore, the less the mobile base moves, the more acceptable it is for use in social environments. Along these lines, virtual fixtures based on nominal gait thresholds are to implemented for the workspace centering control of AssistOn-GAIT. These fixtures are defined with respect to the mobile platform. The controller is implemented such that the mobile platform does not move unless the patient pose comes in contact with the fixtures. If the patient violates the virtual fixture constraint, the penetration distance and direction are calculated and the motion of the mobile base (hence the virtual fixtures) is controlled to eliminate this violation.

Unlike in the mobile haptic interface applications, the mobile platform does not move for every pose change of the patient. In particular, the thresholds are selected such that during straight walking, the mobile base does not have any lateral or rotational movements, since repetitive motions of the pelvis along these DoF can be safely kept within the workspace of the exoskeleton module. However, the thresholds are exceeded if the patient decides to turn or side-step and the mobile base moves accordingly to compensate for these movements.

The workspace centering controller of AssistOn-GAIT operates based on measurements provided by the exoskeleton module. Note that, only relative measurements between the mobile base and the exoskeleton module are required for the implementation of the controller. During walking, the pelvis-hip complex has complex translational and rotational movements; however, the vertical projection of these motions on the floor is of interest for the control of the mobile base. In particular, relevant movements are caused by the rotation of the pelvis in the transverse plane, flexion/extension of the hips in the sagittal plane, forwards/backwards pelvic displacement along the direction of walking and lateral pelvic displacement during walking or side stepping. Thanks to the encoders installed on the LPAM and 3$\underline{R}$RP mechanisms of the exoskeleton module, relative configuration of left and right hip joints, as well as the lateral position of the pelvis can be calculated in real-time. These measurements are used for the workspace centering control, as shown in Figure 5.8. In the figure, the direction of forward walking is denoted

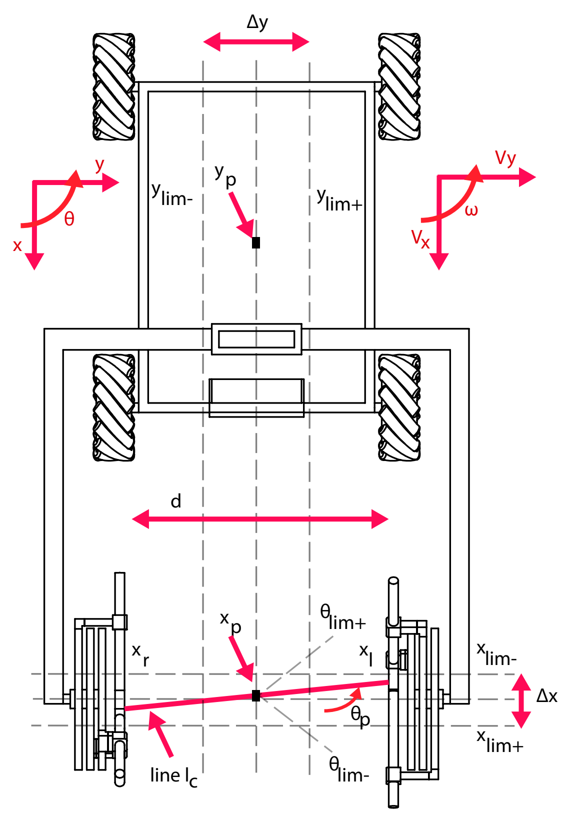

Figure 5.8: Determination of patient pose through positions measured by the exoskeleton module

by the x-axis, the direction of sidestepping is denoted by the y-axis and axis of rotation for turning is given by the z-axis.

The 3$\underline{\text{R}}$RP mechanisms and LPAM have corresponding physical workspace limits and it is imperative that the controller can keep the patient within these limits, by moving the mobile base accordingly. For this purpose, along the x axis, the estimations x_r and x_l are available, which represent the relevant coordinates of the right and left hip joints, respectively. Along the y axis, LPAM directly measures the lateral displacement of the pelvic joint. Based on these measurements, the relative x_p displacement of the patient along x axis and the heading angle θ_p of the patient with the x-axis can be calculated using x_r and x_l, while the relative y_p displacement of the patient along y axis corresponds to the LPAM measurement.

5.4 Experimental Verification of Holonomic Mobile Base Performance

For experimental verification, a healthy volunteer (28 years old male) was connected to AssistOn-GAIT and asked to walk naturally along a predefined path, while avoiding obstacles as depicted in Figure 5.9. For this experiment physical obstacles have been used, even though AssistOn-GAIT can also generate virtual fixtures as demonstrated in [15]. The volunteer was asked to start walking forwards, then stop, sidestep to avoid the obstacle and then turn along a defined path. The actuators of the mobile base and workspace centering controller were active during the trials, along with the position sensors attached to the exoskeleton module. The goal of the workspace centering controller is to ensuring that the displacements/angles of the pelvis-hip complex remain within pre-defined limits of the exoskeleton workspace. Several tests were carried out and a representative trajectory of each of the task phase is presented in Figure 5.10.

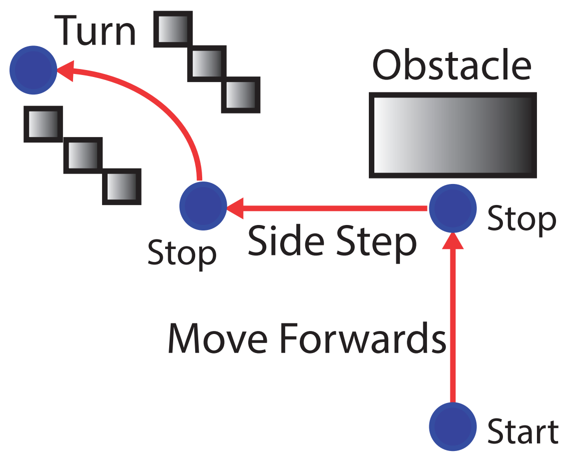

Figure 5.9: Experiment protocol

In Figure 5.10, the first row depicts left hip displacement trajectories during forward walking with virtual fixtures set at ±10 mm, the second row presents lateral displacements trajectories of the volunteer during side stepping with the virtual fixtures set at ±35 mm and the third row plots the pelvic rotation angle trajectories during turning with a virtual fixture set to 8°. Success of the experiment is evaluated by studying two factors: the exoskeleton must not reach the limit of its physical workspace and the mobile base should properly follow the human and not lead or lag in its motion.

It can be observed from the Figure 5.10 that the volunteer can move (forwards, sideways) and turn freely within the virtual fixtures without invoking any mobile base movements. Furthermore, it can be observed that the virtual fixtures are not violated (more than some predetermined margins), indicating that the mobile base can interfere in a timely manner to ensure the desired level of performance.

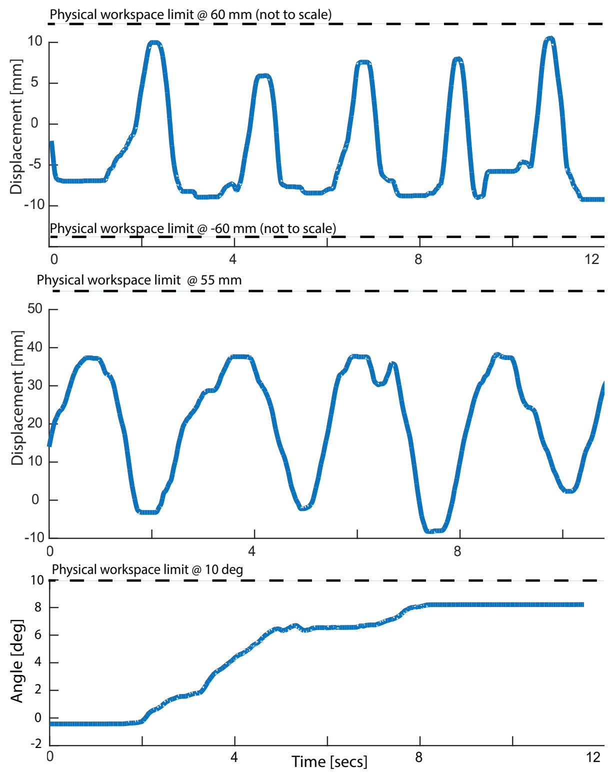

Figure 5.10: Experiment results: (a) Left hip displacement trajectories during forward walking with virtual fixtures set at ±10 mm, (b) Lateral human displacement trajectories during side stepping with virtual fixtures set at ±35 mm and (c) Pelvic rotation angle trajectories during turning with a virtual fixture set to 8°.

5.4.1 Discussion

In this chapter, we have presented the series elastic as well as workspace centering control of the Holonomic Mobile Base, together with their experimental verification. The advantages/disadvantages of both techniques has been presented. The series elastic actuation allows measurement of interaction forces and is able to impose haptic guidance to the connected patient. However, it suffers from problems such as low bandwidth and the requirement of moving the mobile base even when the connected human performs little motion. To overcome these drawbacks we have proposed the workspace centering control.

We have argued that the redundancy between the mobile base and the exoskeleton module is crucial to ensure best possible force rendering performance and capitalized on the redundant kinematics to control the movements of the mobile base through position measurements of the exoskeleton module. The proposed workspace centering controller ensures that the workspace limits of the exoskeleton module are not reached, resulting in an unlimited overground training workspace for AssistOn-GAIT. Furthermore, the workspace centering controller also serves a crucial role in dynamic decoupling between the mobile base and the exoskeleton module.

Chapter VI

6 Dynamic Body Weight Support System

Body weight support (BWS) systems are an indispensable component of robot assisted gait rehabilitation. It has been clinically shown that gait rehabilitation can be made significantly more effective, when a percentage of patients' weight is supported by a BWS system, as compared to the case when patients have to bear their full weight [55–57]. Furthermore, BWS is essential during robot assisted therapies, to ensure safety and to prevent falls.

In this chapter, we present a series elastic active weight support and inertia compensation system for use in gait rehabilitation. The system is capable of providing dynamic weight support to patients while walking. In addition, it can provide compensation for the inertial forces caused by the vertical movements of the human body. Compensation of inertial forces has been largely ignored in the literature, even though these forces can cause significant deviations in the unloading force, specially when the support force is low and walking speed is fast. Furthermore, without such compensation, the added inertia due to the device may interfere with natural gait. Gravity-Assist aims to overcome these limitations by actively compensating for inertial forces using online measurements of the vertical accelerations. Furthermore, thanks to series elastic actuation, robust force control and low output impedance can be achieved with a relatively low-cost device.

6.1 Body Weight Support in Gait Rehabilitation

Patients with walking disabilities are often unable to support their own weight, due to muscle weakness or paralysis. Consequently, an effective gait rehabilitation system must be capable of fully/partially supporting the weight of the patient. Such systems can help to reduce the force that patients encounter on their legs during walking.

Efficacy of gait rehabilitation therapy while supporting the weight of the patient has been explored by many groups. Experimental results indicate that gait rehabilitation is more effective when the body weight of the patient is partially supported [55–57]. It has been documented that a weight support of no more than 30% of the body weight results in best performance, as high levels of weight support can decrease the activity of muscles in stroke patients. Furthermore, effective training is known to require gradual reduction of the weight support as patients start supporting their own body weight [94]. Lateral balance of patients has also been shown to improve, when BWS is used in gait rehabilitation [58]. Furthermore, BWS is essentially required to ensure safety and to prevent falls during walking. This feature is especially important for overground gait rehabilitation devices (as compared to treadmill based devices) where the risk of falling is greater [1, 59].

BWS systems are found in many gait rehabilitation devices to unload the patient weight during walking and to prevent falls. With respect to weight unloading, these systems can be categorized into i) static systems, ii) passive counterweight based systems, iii) passive elastic spring based systems, and iv) active dynamic systems. Figure 2.1 presents a schematic representation of these categories [1].

Static systems consist of a mechanism that can be set to unload a predefined amount of weight, when the patient is connected to the support system [95]. However, these systems do not ensure equal amount of weight support as the body center of mass (CoM) moves vertically during walking. Especially, when the body moves downwards the harness becomes tighter, restricting the movements of the pelvis. This makes static systems unsuitable for natural walking and make them less effective for gait rehabilitation. *Passive counterweight based* systems use a counterweight to dynamically unload a percentage of the patient's body weight [96]. The counterweight is set to a predefined value and moves vertically as the patients walks, maintaining a constant static unloading force. However, the movement of the counterweight also results in additional inertial forces, which can cause large fluctuations in the support forces, unless actively compensated. *Passive elastic* systems use elastic elements to provide unloading forces based on the tension in these elastic elements [97]. This is advantageous with respect to the counterweight based approach, since these systems do not induce extra inertial forces; however, passive elastic systems cannot ensure that the unloading forces stay constant, as the amount of support varies with the length of the elastic element. Consequently, all of these passive systems are not very effective for body weight support, as the patient does not feel a constant weight unloading during walking, interfering with the natural gait and potentially negatively affecting the efficacy of the therapy.

Active dynamic systems are capable of generating unloading forces dynamically [1]. In particular, these type of systems continually measure the interaction force between the patient and the BWS actuator and based on these measurements, a control system commands the actuator to move in such way that a constant amount of vertical force is felt by the patient, despite the vertical movements of the patient during walking. With a high enough control bandwidth, these systems can provide comfortable weight unloading to promote natural walking [59, 60].

A number of strategies have been employed for connecting the patient to a BWS, including utilization of an overhead system of pulleys or direct connection to the trunk, waist, hip or pelvis of the patient. The overhead pulleys is the most commonly employed method; however, this arrangement tends to produce horizontal forces when patients are not directly under the support system. These forces may pull patients towards the center and introduce balance problems. In particular, the index of lateral stability becomes low with these systems, when the supporting weight is high [67]. Even though overhead pulleys are commonly used and easy to implement, other methods of connection may be preferable to ensure better balance during gait training.

6.2 Inertia Compensation

When a patient is attached to a BWS system, the human body acts as an inertial load. The human weight exerts a static force on the BWS, while the vertical movements of the body during walking cause inertial forces which are proportional to its acceleration. Considering the human mass and natural walking speeds, these inertial forces can become significantly large and hinder the operation of the BWS, by causing large deviations in the interaction force from the desired level unloading. Furthermore, the added inertia of BWS may cause a decrease in the natural frequency of motion, hindering achievement of natural gait [58, 98].

A more natural unloading strategy for gait rehabilitation is to compensate, not only for the partial weight of the patient, but also for the corresponding inertial forces due the compensated mass and the BWS system, such that the patient feels as if his/her total mass has been reduced by a predetermined amount even under dynamic movements. This strategy avoids an improper ratio between the weight and inertial forces rendered to the patient and has the potential to promote natural gait. For natural motion, the percentage of inertia compensation is set to the same level as the percentage of weight unload.

While the compensation for patient/device weight and parasitic effects, such as friction and stiction, can be robustly achieved through a force controller, the inertia compensation is more challenging due to the stability issues it presents when used in a feedback control loop [58, 98]. To alleviate these stability problems, commonly employed inertia compensation approaches rely on an extra sensor to detect the instantaneous accelerations of the compensated inertia and a model-based feed-forward compensator. In particular, after presenting the coupled stability limitations of closed-loop inertia compensation approaches, [98] proposes an *emulated inertia compensation* scheme that utilizes the filtered acceleration measurements to approximately compensate for the limb and exoskeleton inertia for a lower-extremity exoskeleton. This study also provides evidence that such an inertia compensation scheme can improve the natural frequency of lower-limb swing movements.

6.3 Force Control and Series Elastic Actuation for Dynamic Body Weight Support System

Force control is one of the crucial components of an active BWS. In the literature, stiff commercial force sensors are commonly employed in admittance control schemes. Due to the non-collocation between the force sensor and the actuator, when stiff force sensors are used, the maximum gain that the force controller can utilize is severely limited by stability constraints. Low controller gains are undesirable, since they limit the robustness of the controller and the accuracy of force control. Furthermore, with stiff force sensors, any sudden impacts from the environment are directly passed to the controller with potentially damaging effects.

Series Elastic Actuation (SEA) is a technique proposed to alleviate these issues by deliberately introducing a compliant element between the actuator and the environment. This compliant element has a stiffness that is orders of magnitude less than the stiffness of a force sensor; hence, SEA allows for higher controller gains to be employed for robust and accurate force control. Force is determined by measuring the deflection of the compliant element, while force control can be implemented through robust motion control of the compliant element. Mechanically, SEA possesses a low output impedance at frequencies higher than its control bandwidth, enabling these systems to absorb any sudden impacts from the environment. In addition to low output impedance, SEA features active backdriveability within its control bandwidth, which makes this technique suitable for applications requiring physical human-robot interaction. SEA can be implemented at low costs, since neither high quality drive trains/power transmission elements, nor expensive force sensors/signal conditioners are required for these systems. A potential disadvantage of SEA is its limited force control bandwidth; however, bandwidth limitation is not of high concern for rehabilitation robotics where the patient movements are relatively slow.

Compensation for inertial forces due to patient has been mostly ignored in

BWS systems available for gait rehabilitation. However, considering that the amplitude of vertical CoM displacements is about 49.8 ± 10.3 mm at natural gait cycles [69], the inertial forces can account up to 25% of the gravitational loads.

Similar to ZeroG and FLOAT, Gravity-Assist is a series elastic active BWS system that can be used to unload patients' weight during gait rehabilitation. Unlike ZeroG and FLOAT, Gravity-Assist actively compensates for the inertial forces utilizing the emulated inertia compensation scheme [98]. Furthermore, the device connects to the trunk to feature improved lateral stability during walking.

6.4 Gravity-Assist Dynamic Body Weight Support System

Gravity-Assist is designed to satisfy the following task specifications. The active BWS system should

- dynamically compensate for inertial forces of the patient's body, in addition to unloading of patient's weight,
- not interfere with patient balance,
- ensure safety against falls,
- allow for unrestricted pelvic movements,
- provide an ergonomic and comfortable support,
- be able to lift patients from a sitting position to initiate therapies, and
- enable low cost implementations.

6.4.1 Design

To satisfy the task specifications, Gravity-Assist features a single DoF series elastic BWS system that attaches to the trunk of a patient. A connection to the trunk is preferred as it does not interfere with pelvic movements or the balance of patients. SEA ensures robust, high fidelity control of interaction forces, while simultaneously ensuring active backdriveability and a low output impedance. SEA also enables low cost implementations of the system, through use of low cost power transmission elements, robust motion controllers and position sensors. The system features appropriate stroke and output force to actively assist patients during sit-to-stand tasks. Finally, the bandwidth of the SEA is designed to be significantly higher than the 1 Hz bandwidth of natural human walking [38].

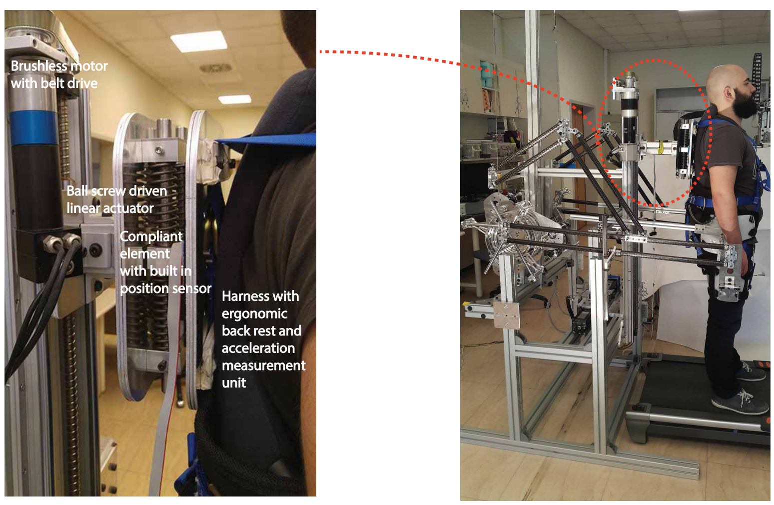

Figure 6.1: Gravity-Assist attached to a volunteer

Gravity-Assist consists of three modules: series elastic element, motion controlled linear actuator, and harness and acceleration measurement unit as shown in Figure 6.1.

Linear Actuator

The linear actuator consists of a low friction ball screw mechanism, whose screw is actuated by a velocity controlled brushless DC motor connected through a belt drive. The actuator is selected such it can continuously support the weight of a patient and has a stroke of 1000 mm to enable lifting the patient from a sitting position. The velocity controller is implemented on the motor driver hardware.

Series Elastic Element

The series elastic element consists of linear compression springs sandwiched between two aluminum plates, one attached to the end-effector of the linear actuator and the other attached to the harness. The deflection of the springs are measured by an optical encoder that possesses 8000 count per inch resolution under quadrature decoding. Given the stiffness of the spring, the interaction force between the two plates can be estimated.

Harness and Acceleration Measurement Unit

The patient connects to the series elastic element through a harness that is fitted with an ergonomic back support for proper posture and comfort. Vertical force exerted by the human body is transmitted through the harness to the series elastic element, where it is measured as spring deflection. Passive movements of the harness within limited ranges are enabled (except along the vertical direction) to promote patient comfort.

The acceleration measurement unit consists of a three axes low-g MEMS accelerometer with an internal sampling frequency of 11 kHz. The accelerometer is attached to the patient using a chest strap. The acceleration measurements are low pass filtered.

6.4.2 Control

A real-time cascaded controller is implemented for SEA as shown in Figure 6.2. In the figure P_τ, I_τ represent the gains of the outer PI force controller,

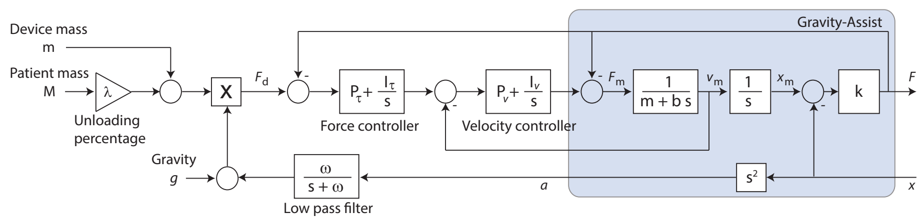

Figure 6.2: Cascaded force controller with emulated inertia compensation

P_v, I_v represents the gain of the inner PI velocity controller, m, b represents the inertia and damping of the motor, k is the stiffness of the compliant element, ω is the cutoff frequency for the low pass filter, M, m represent the mass of the human and end effector respectively. The signal F_d represents the desired force, F_m is the motor force, F is the actuator output force, x is the displacement caused by human. The cascaded controller consists of an inner robust velocity control loop and an outer force control loop. The inner loop of the control structure deals with imperfections, such as friction and stiction of the linear actuator, rendering the system into an ideal velocity source. The velocity controller is implemented in hardware on the motor driver with fast control rate of 20 kHz. The outer force control loop is implemented at 1 kHz for high fidelity force control. This loop imposes the desired level of support forces as determined by the therapist. The cascaded force controller [99] is augmented with emulated inertia compensation scheme as proposed in [98]. In particular, inertial forces to be compensated for are estimated based on low-pass filtered real-time acceleration measurements and the pre-determined mass of the device and the patient. Then, these inertial force estimates are provided to the force controller as a reference, in addition to the force reference used for weight unloading based a predetermined percentage ($\lambda\%$) of the patient weight set according to the therapy requirements.

For safety, in addition to emergency stops, limits on the maximum speed and minimum position of the linear actuator have been implemented. In case the patient falls or is unable to support his/her own weight, the linear actuator holds the patient and slowly moves down to a comfortable height, where a stool can be placed under the patient. The system also allows for connecting of patients at a sitting position and patients to be slowly raised to a standing position.

6.5 Experimental Characterization of Gravity-Assist

Gravity-Assist has been experimentally characterized and the technical specifications of the prototype are summarized in Table 6.1.

Figure 6.3 presents the small, moderate and large force control bandwidths of the device. With 4 Hz large force control bandwidth, the device is evaluated to be significantly faster than natural human gait cycle taking place at 1 Hz.

Figure 6.4 depicts force tracking performance of the device for a chirp force reference input, for which an RMS error of 1.34% can be reported for

Table 6.1: Technical specifications of GRAVITY-ASSIST

Parameter	Value
Stroke	1 m
Output force	600 N
Force sensing resolution	0.3 N
Velocity control bandwidth	12 Hz
Force control bandwidth	9.5 Hz (20 N – low force)
	6.5 Hz (250 N – moderate force)
	4.0 Hz (500 N – high force)

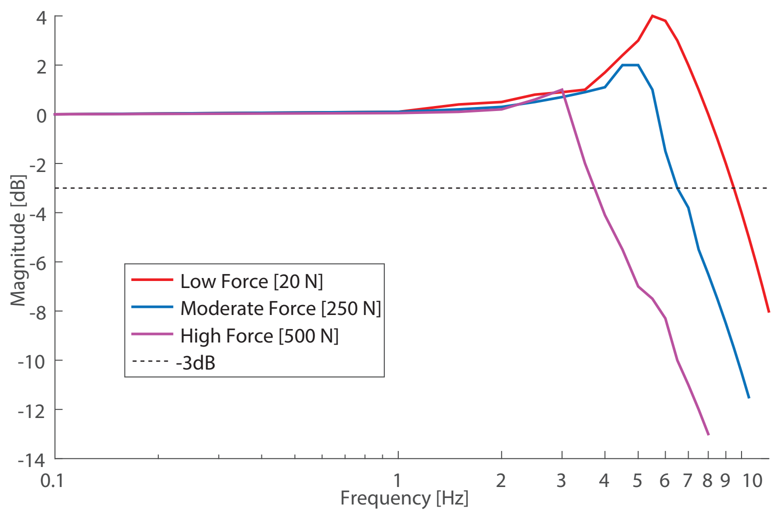

Figure 6.3: Experimental characterization of the small, moderate and large force control bandwidths of GRAVITY-ASSIST

frequencies up to 5 Hz.

6.6 Evaluation of Inertia Compensation Performance

We have experimentally evaluated the effect of inertia compensation on the assistance provided. For the trials, we mounted GRAVITY-ASSIST on a treadmill. A 26 years-old healthy volunteer was connected to the device. The volunteer signed an informed consent form approved by the IRB of Sabanci University before taking part in the experiments.

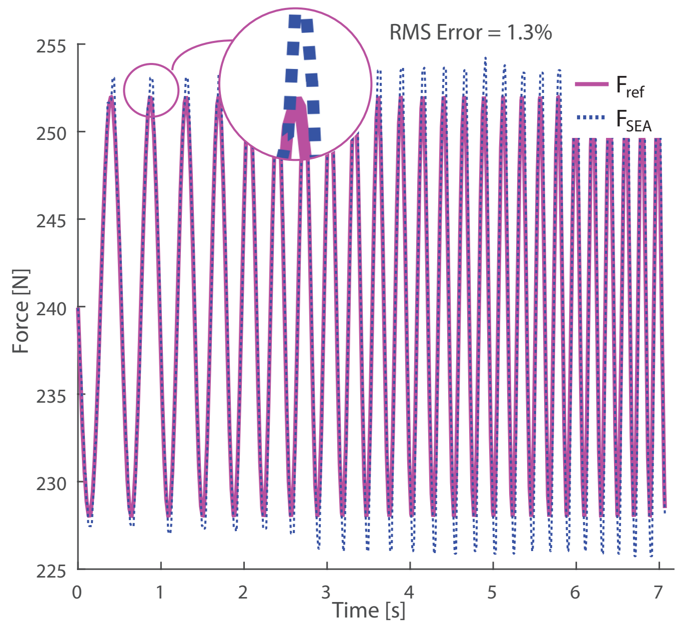

Figure 6.4: Force tracking performance of GRAVITY-ASSIST for a chirp force reference input

The experimental protocol consisted of attaching the device to the patient at a sitting position, raising him to a standing posture and asking him to walk forward at a natural pace. During all trials, the control system was set to actively support 50% of the volunteer's 64 kg weight. Three conditions have been tested for inertia compensation: 0% inertia compensation (simple gravity compensation case), 50% inertia compensation (to match the percentage of weight compensation) and 100% inertia compensation. After each experiment, the volunteer was also asked to report the comfort level of the assisted gait. The order of trials was randomized. The vertical acceleration of the volunteer was measured using the accelerometer attached to his chest and a cutoff frequency 3 Hz was used for low-pass filtering these measurements.

The upper row of Figure 6.5 presents representative plots depicting reference assistance forces and measured interaction forces between the device and the volunteer, for weight unloading with and without inertia compensation. In particular, the inertia compensation is turned off in Figure 6.5(a), where a constant force is rendered by the device to compensate for 50% of the volunteer's weight. In this plot, one can observe relatively large deviations of the measured interaction force from the reference due to inertial force contributions, which are as large as 15% of the assistance forces. In Figure 6.5(b) and (c), 50% and 100% of the inertial forces are compensated for by the controller. As expected, the difference between the reference assistance forces and the measured interaction forces decreases as higher percentage of inertial forces are compensated, since the main disturbance acting on the force control system consists of the unaccounted inertial forces.

The lower row of Figure 6.5 presents the interaction forces between the ground and the volunteer. In Figure 6.5(e), the ground interaction forces include the inertial forces due the mass of the volunteer, while in Figure 6.5(f) the ground interaction forces are kept close to the constant value of uncompensated weight, as almost all of the inertial forces are successfully compensated for. The 50% compensation case is presented in Figure 6.5(f), where the ground interaction forces closely match the calculated interaction forces for a fictitious agent with 50% of the volunteer's mass. This case is interesting, since ensuring a meaningful ratio between the weight and inertial forces may promote patients to achieve a more natural gait.

Interview with the volunteer after the trials indicates that weight support without inertial compensation results in a gait that feels unbalanced and uncomfortable, the case with full inertia compensation lacks the dynamics of the gait, while the 50% compensation feels relatively more natural.

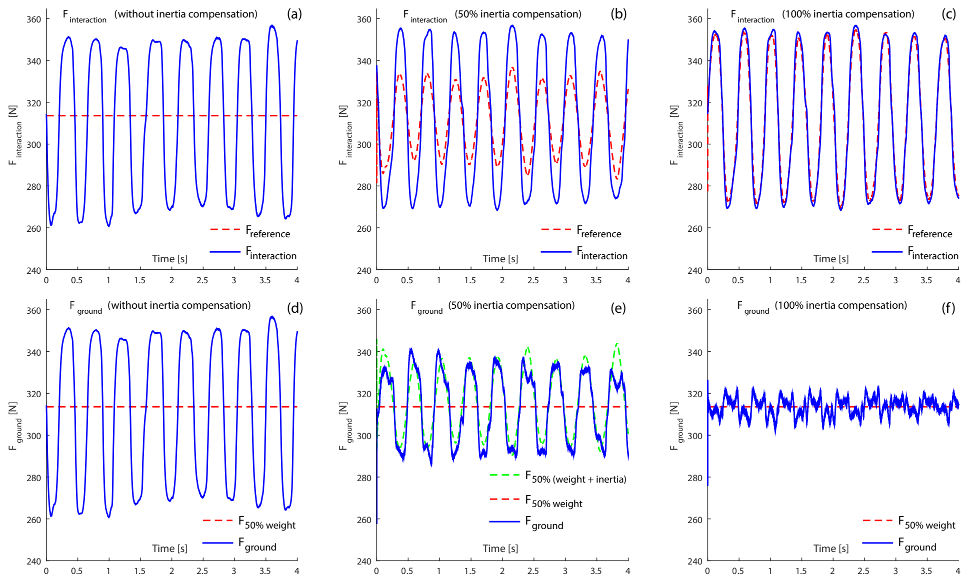

Figure 6.5: Forces between volunteer and device (a) without inertia compensation, (b) with 50% inertia compensation, and (b) with 100% inertia compensation. Forces between volunteer and ground (d) without inertia compensation, (e) with 50% inertia compensation, and (f) with 100% inertia compensation.

6.6.1 Discussion

In this chapter, a series elastic BWS and inertia compensation system has been presented and experimentally characterized. The inertia compensation has been performed in a feed-forward manner based on online acceleration measurements taken from the trunk of the patient, while a cascaded force-motion controller has been used for force control for SEA. Initial experiments performed on a healthy volunteer indicate that deviations from desired interaction forces can be significantly reduced when inertia compensation is utilized. Furthermore, the volunteer states that the gait feels more natural when the inertia compensation matches the weight unloading.

Chapter VII

7 Performance Characterization and Human Subject Experimentation

The performance of AssistOn-GAIT has been experimentally characterized and presented in this section.

7.1 Workspace Characterization of Pelvis-Hip Exoskeleton

The workspace limits of individual modules of AssistOn-GAIT have been presented earlier. Each individual human being however has different body proportions based on which they can utilize the available workspace. Furthermore, the mechanical design of the system can present some constraints. In this section we characterize the practically available limits along each of the 7 pelvis-hip degrees of freedom. It involves using a healthy human volunteer to identify the limits of movement along each degree of freedom. In each case it has been shown that the workspace is significantly larger than human range of motion required for gait therapy. The characterization procedure and points where measurements haver been made is shown in Figures 7.1-7.5. Results are listed in Table 4.1.

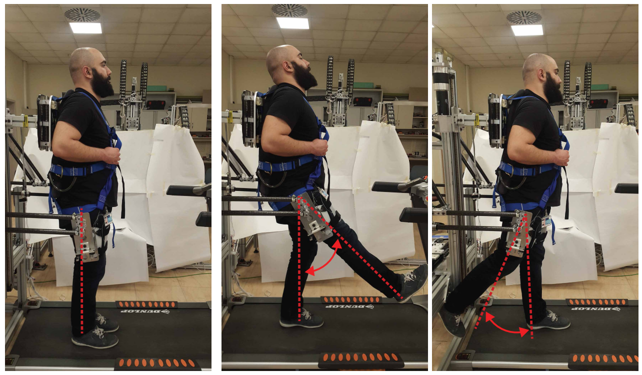

Figure 7.1: Method for characterizing available workspace for hip flexion-extension

7.2 Trajectory Tracking Performance of Pelvis-Hip Exoskeleton

Under impedance control, the EXO is able to track desired trajectories within its workspace with a high precision. To characterize the error in trajectory tracking, sinusoidal reference tracking was performed along the three DoF of the EXO and results are shown in Figure 7.6. A circular trajectory was generated in the sagittal plane, and the results alongwith RMS erros are shown in Figure 7.7. Results indicate that the EXO is able to track desired trajectories in the sagittal plane with a very high accuracy.

7.3 Passive Backdriveability of Pelvis-Hip Exoskeleton

The EXO is a passively backdriveable system and its connection to the human (motion transmitter and human connection) have been designed to be light and feature low weight and inertia. Thus, the design features the minimum impedance mode interaction forces and torques low. The backdriveability

Figure 7.2: Method for characterizing available workspace for lateral pelvic displacement

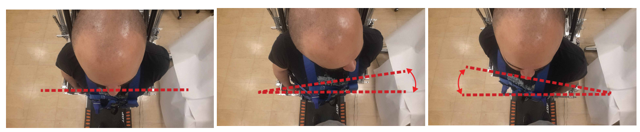

Figure 7.3: Method for characterizing available workspace for pelvic rotation

has been measured by noting the force required to move the EXO along the sagittal axis, longitudinal axis and torque required to move it about the horizontal axis. Current was input to the motors in small increasing steps until the end effector started moving. The value of this current in combination with the motor torque constants and Jacobian Transpose was used to calculate the forces and torques. Backdriveability values have been calculated as 5.5 N along the sagittal axis, 6 N along the longitudnal axis and 0.385 Nm about the horizontal axis.

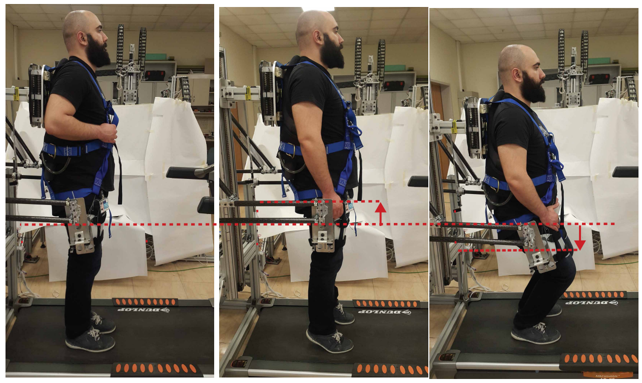

Figure 7.4: Method for characterizing available workspace for vertical pelvic displacement

Figure 7.5: Method for characterizing available workspace for pelvic tilt

7.4 Impedance Rendering Performance of Pelvis-Hip Exoskeleton

The EXO is impedance controlled and allows rendering of desired impedance values. To characterize the rendering accuracy, a stiffness value of 6 N/mm was rendered and checked with a force sensor. Characterization results and errors are shown in Figure 7.8.

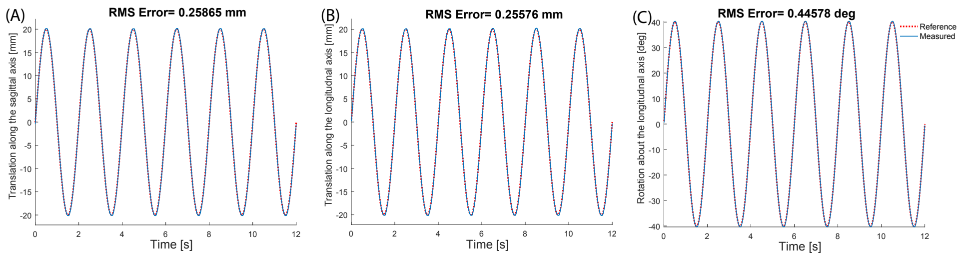

Figure 7.6: EXO RMS Error characterization for sinusoidal reference tracking at 1 Hz (A)Translation along the sagittal axis (B)Translation along the longitudinal axis(C) Rotation about the horizontal axis

7.5 Donning Time of AssistOn-GAIT

The time required to connect a patient to a rehabilitation robot is very important. A robot that requires a long donning time would have limited usability and also be a cause of frustration for patients who are already sick or in pain. This time is mainly required for aligning human joints with the exoskeleton and may require measurements of limbs of the patients body to accommodate for different body sizes. The donning time of a very few rehabilitation robots has been published in literature. The time for HAL robot is 15 to 20 minutes, for PAM/POGO it is 30 minutes and for Lokomat it is 20 minutes for the first session. One of the lowest donning times has been reported by LOPES II [39] which is 10 to 14 min for first-time training and 5 to 8 min for recurring training.

AssistOn-GAIT has been designed such that it features a very low donning time. The EXO consists of two passively back driveable 3$\underline{R}$RP mechanism which align automatically to the pelvis/hip of te patient when the straps are tightened. The BWS is actively backdriveable and automatically adjusts to the trunk height when straps are tightened (the system operates

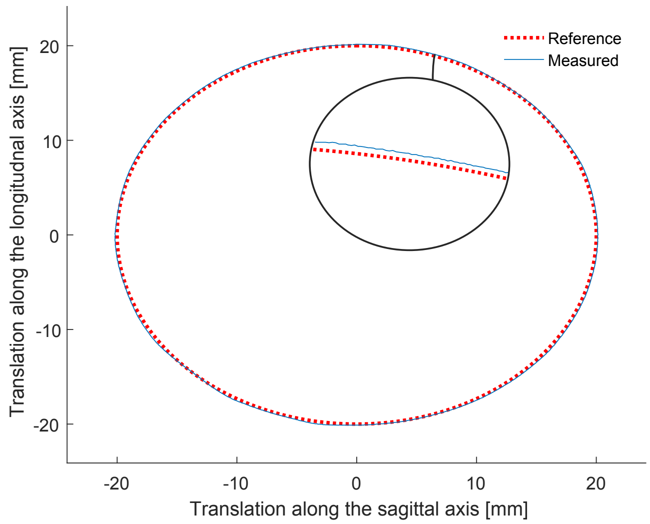

Figure 7.7: Circular trajectory tracking for the EXO in the sagittal plane

in minimum impedance mode). Only the LPAM requires measurements of patient hip height and width in order to adjust its vertical and horizontal position. Therefore, the overall system requires very less time for connecting the patient and only requires a one time measurement of the hip width and height.

For characterizing the donning time, a volunteer was connected to the system and the time was noted. The first step consists of measuring the patients hip height and width, and adjusting the LPAM connection accordingly. Then the patient is asked to step onto the treadmill and the BWS straps are connected, followed by the LPAM strap and EXO straps. Repeated experiments were performed and it has been determined that the average donning time of AssistOn-GAIT is 5 minutes for the first session and 1 minute for subsequent sessions with the same patient.

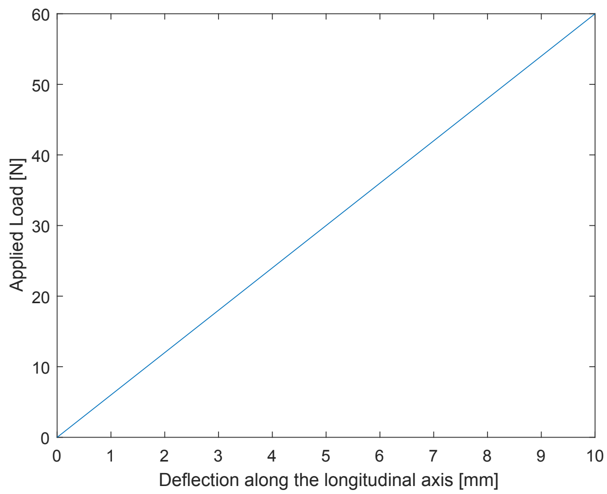

Figure 7.8: Rendering a 6 N/mm stiffness along the longitudinal axis

7.6 Characterization of Series Elastic Body Weight Support and Lateral Pelvic Actuation Modules

The BWS and LPAM have a similar design principle and utilize similar hardware. Therefore, the characterization of only the BWS is performed and is presented in Chapter-IV.

7.7 Human Subject Experiments to Determine Effectiveness of Pelvic Guidance

Human experiments with the BWS and Workspace Centering have been performed presented in Chapter-III and Chapter-IV. In this section, experiments with the force rendering capabilities of the EXO are presented. The experiments have been performed on a 26 years old healthy male volunteer. Informed consent was obtained from the subject in line with the rules set by

Sabanci University.

When operated in the MI (minimum impedance) mode, the EXO is passively backdriveable and the human subject is asked to walk naturally. Data relating to pelvis-hip translations and rotations is collected. Representative data for hip flexion/extension is shown in Figure 7.9. Data collection helps with analysis of gait and programming for haptic guidance of pelvis-hip trajectories.

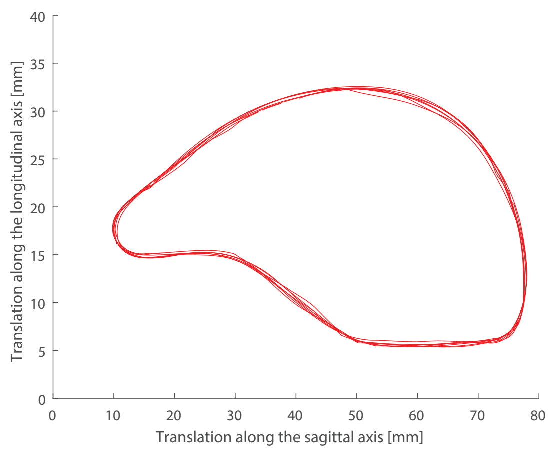

Figure 7.9: Plot showing a closed pelvis/hip trajectory collected using the EXO

The ability of the EXO to render desired trajectories along its DoFs under impedance control is the most important feature of the EXO. Desired stiffness along the trajectory serves as a haptic guidance for the human connected to the EXO. Therefore, pelvis-hip movement guidance can be imposed on the patient. This guidance may be helpful against compensatory movements of the pelvis. In this experiment a 26 years old healthy male volunteer was connected to the EXO and asked to walk naturally while the EXO was operated in MI (minimum impedance mode).

Data was collected during this phase, which indicated the pelvis-hip trajectory for the subject in the sagittal plane. An average trajectory was then extracted from the data and modeled as a Non-uniform rational basis spline (NURBS) curve, to represent the target trajectory. Two offset NURBS curves were generated, one on the outer side of the trajectory to represent the outer virtual wall, and one on the inner side of the curve to represent the inner virtual wall, thus creating a virtual tunnel between the walls. The location of the pelvis-hip is measured and its distance from the target trajectory is calculated in real time using a closest point algorithm. The pelvis-hip is free to move inside the tunnel, however, in case any of the virtual walls is penetrated gentle haptic forces are applied in the opposite direction to guide the pelvis-hip back within the virtual tunnel.

Results of the experiment are shown in Figure 7.10 and 7.11.

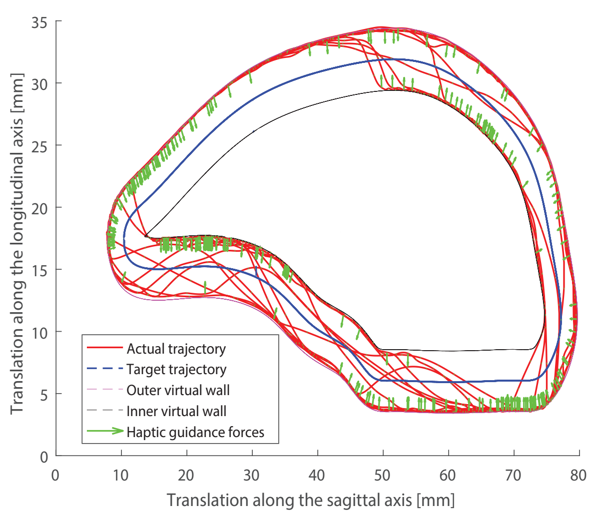

Figure 7.10: Pelvic guidance using a wide virtual tunnel

The experiment is started and the volunteer is asked to walk on the

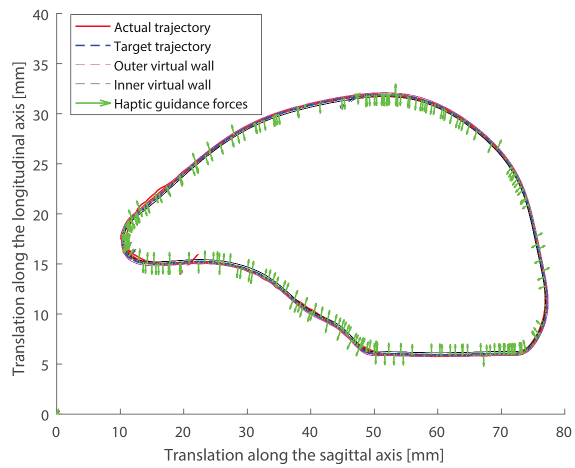

Figure 7.11: Pelvic guidance using a narrow virtual tunnel

treadmill naturally but make small attempt to deviate from his normal style of walking. Initially a wide virtual tunnel (8 mm) is generated. The pelvis is free to move within the virtual fixtures. During walking, in case the pelvis deviates from the designated trajectory, gentle haptic forces guide the pelvis and keep it inside the desired trajectory defined by the virtual fixtures. In the second experiment, a narrow virtual tunnel (1 mm) is generated. This time the haptic forces make sure that a trajectory very close to the desired trajectory is followed.

7.8 Human Subject Experiments to Determine Gait Patterns with and without AssistOn-GAIT

AssistOn-GAIT has been designed not to restrict any pelvic degree of freedom. The EXO module has been designed to be passively backdriveable using low friction capstan transmission. Thus when a human is connected to the device and the device operates in minimum impedance mode, the gait pattern of the human stays very close to the natural pattern. To verify this claim, experiments with human subjects have been conducted where human gait is analyzed using a motion capture system. The joint angles/displacements are measured when the human walks with and without the exoskeleton, and a comparison has been made. Details of the experimental setup and results are discussed in the following paragraphs.

7.8.1 Experimental Setup

The Xsens full body measurement has been used for motion capture and gait analysis. The system consists of a total of 17 wireless motion sensors attached to the volunteers body(Figure 7.12). The sensors are able to capture motion data at a rate of 60 Hz and transmit it to a host computer where a software uses the data to extract joint angles, joint displacements and create real time 3D animation of the walking. Data related to joint angles, segment positions and body center of mass is displayed and recorded [100].

7.8.2 Protocol for Human Subject Experiments

Human subject experiments have been carried out with a total of 8 healthy subjects (04 female, 04 male) whose details are mentioned in Table 7.1.

Table 7.1: Human Subject Details for Multi Subject Experiments

Subject	Sex	Age (years)
Subject 1	Male	24
Subject 2	Male	25
Subject 3	Female	25
Subject 4	Male	23
Subject 5	Male	25
Subject 6	Female	23
Subject 7	Female	35
Subject 8	Female	29

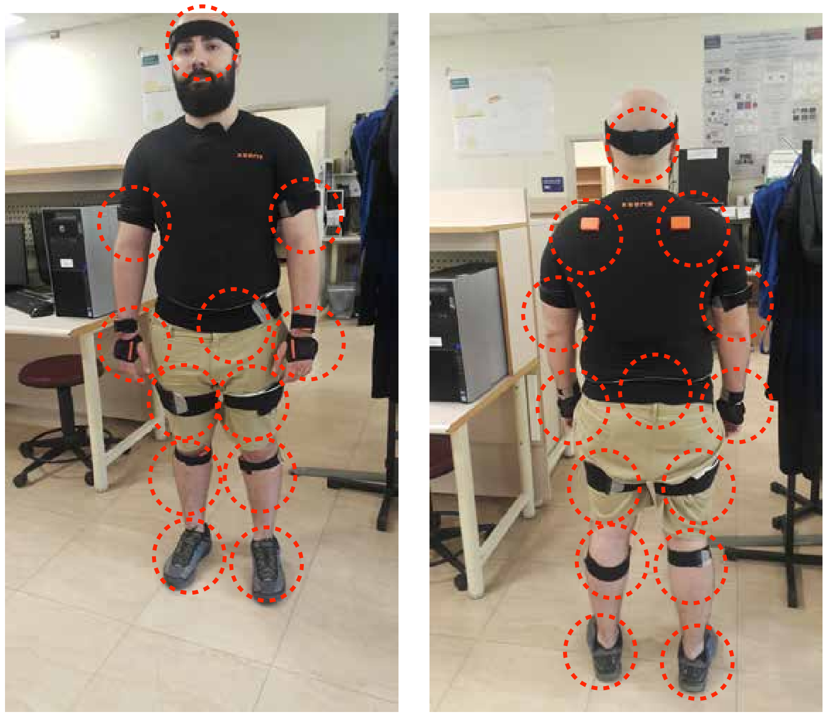

Figure 7.12: A volunteer wearing Xsens sensors (dotted lines show locations of sensors)

Initially the Xsens body measurement system is worn by the subjects and calibrated. Next, the subjects are asked to walk freely on the treadmill at their self selected speed, while their gait pattern is captured. Using the Xsens software, hip joint angle, knee joint angle and vertical displacement of the Center of Mass is recorded. In the second step, the subjects are asked to walk on the treadmill with the EXO connected and self selected step length, while the same body parameters are recorded. The EXO operates in minimum impedance mode and does not assist or resist the patients actively. The only resistance felt by the patients is due to the friction and inertia of the EXO. It is ensured that the subjects walk at the same part of the treadmill during the experiment (walking close to the edge of the treadmill may result in involuntary reduction of step length). Finally a comparison is made indicating how much the gait pattern deviates when the subject walks with and without the EXO.

7.8.3 Results and Discussion for Human Subject Experiments

Results of the experiments are shown in this sub section. Knee joint angles, Hip joint angles and Center of Mass (COM) Position averaged for all subjects with and without the exoskeleton are shown in Figures 7.13, 7.14 and 7.15.

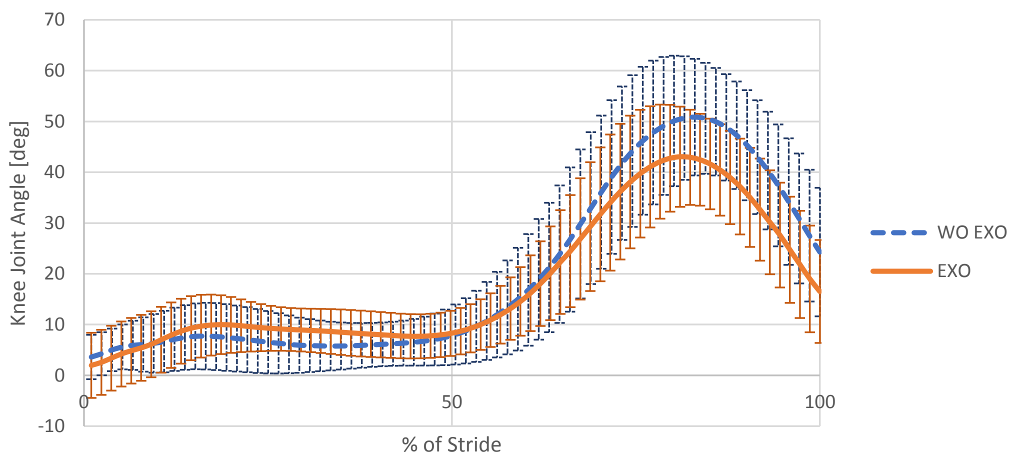

Figure 7.13: Averaged Knee Joint Angle Data for 8 Healthy Human Subjects

The dotted lines show the average trajectory followed by knee joing angle, hip joint angle and COM position when subjects walk freely without wearing the EXO, while the solid lines show the average trajectory followed by knee joing angle, hip joint angle and COM position when the subjects walk connected to the EXO. Error bars showing one standard deviation of the data collected from each subject are also shown. Dotted error bars are for the data without wearing exoskeleton while the solid error bars are for data with exoskeleton. The region covered by error bars represents inter subject variation in the trajectories within one standard deviation.

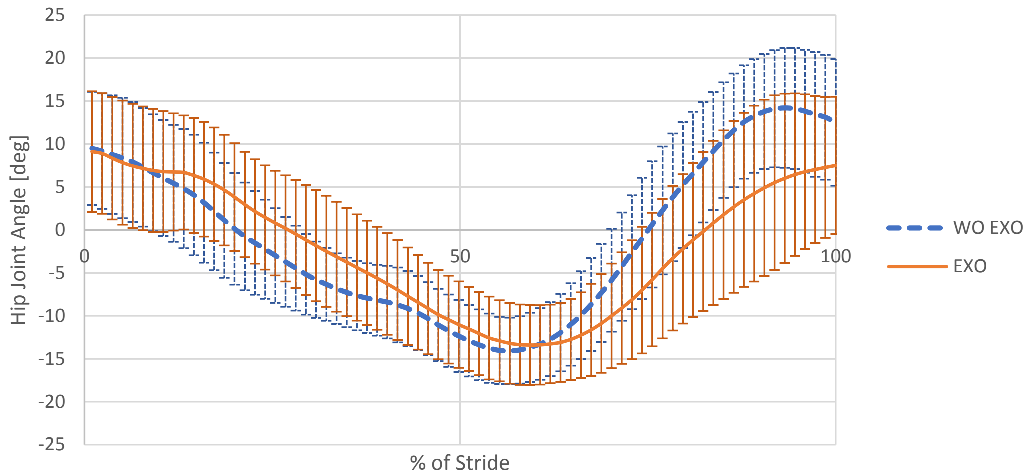

Figure 7.14: Averaged Hip Joint Angle Data for 8 Healthy Human Subjects

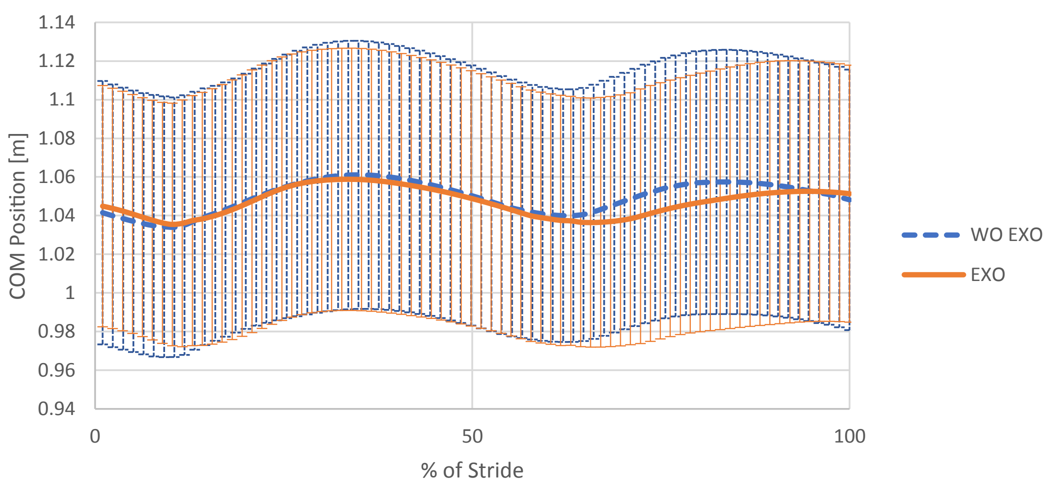

Figure 7.15: Averaged Center of Mass (COM) Position Data for 8 Healthy Human Subjects

7.8.4 Conclusions of Human Subject Experiments

Results for knee joint angle show that both trajectories (with and without EXO) closely follow each other in the stance phase. During the swing phase a small reduction in angle is noted when subjects walk with EXO as compared to walking without EXO. It is important to note that this reduction does not exceed the inter subject variation.

For the hip joint angle, the trajectories show a small increase in joint angle when walking with EXO in the stance phase, while in the swing phase walking with EXO shows a small reduction in angle. These deviations also do not exceed the inter subject variation.

For the COM position, negligible difference is noted between the trajectories with and without EXO.

Overall, it is concluded that the EXO is highly passively backdriveable and causes minimal hinderance to natural gait of healthy human subjects, even when it is not actively controlled.

Chapter VI

8 Conclusions and Future Work

In this chapter the dissertation is concluded by highlighting its major points of interest. Furthermore future works that will follow from this dissertation are listed.

8.1 Conclusions and Discussion

In this dissertation:

- The first full prototype of AssistOn-GAIT, a novel robotic rehabilitation device designed to administer overground gait and balance rehabilitation training to adult patients with gait and balance disorders, has been presented.
- The device has a modular design and consists of three modules, each of which can be operated independently. In particular, the device consists of a Holonomic Mobile Base, a self-aligning active Pelvis-Hip Exoskeleton and an active Body Weight Support System.
- A working prototype of the device has been developed and its characterization and experiments have been performed.

- By performing experiments on human subjects, it has been shown that AssistOn-GAIT provides minimal hindrance to natural walking when operated in minimal impedance mode and is able to guide pelvis-hip movements as desired by the rehabilitation protocol.

8.2 Future Work

AssistOn-GAIT is a potent device that is capable of not only rehabilitation but gait analysis also. Gait analysis is a very important area of research whose understanding is imperative for effective gait rehabilitation. Analysis of gait includes joint displacements, rotations and factors like metabolic cost of walking in a certain way [101]. With the prototype designed, controlled and characterized the next phase would be detailed experimental verification of how the device can be used for gait analysis as well as for rehabilitation. For this purpose, in the first phase, a set of experiments have been designed which are explained below.

8.2.1 Verification of Data Measurement with Motion Capture

The position sensors of the EXO are located away from the human body (due to the fact that motion is transmitted using the Motion Transmitter Mechanism). Optical tracking can be performed to measure the accuracy of the EXO sensors in measuring pelvis-hip angles.

Experiment

Place optical markers on human joints. Use optical tracking system to calculate joint angles. Compare with data obtained from sensors. Calculate errors in measurement for pelvic tilt, pelvic rotation, hip flexion, hip extension, vertical pelvic displacement, lateral pelvic displacement

8.2.2 Evaluation of Importance of Each Pelvic Degree of Freedom

Many rehabilitation devices restrict certain pelvic movements. Furthermore, misalignment between exoskeleton axes and joint axes results in restriction of pelvic movements. This causes gait to deviate from Free Walking and reduce the effectiveness of rehabilitation exercises.

Experiment

Make the connected human walk connected to AssistOn-GAIT . Then restrict a certain DoF of the pelvis and note the effects. Measure ROM (Range of Motion) of hip joint, self selected walking speed, step length and stride length.

8.2.3 Evaluation of Importance of Arm Swing

Arm swinging is an important part of gait and helps in reducing angular momentum (Bruijn,2010), reduces energy expenditure and enhances recovery from gait instability. Many rehabilitation devices restrict arm swinging. Limited research has been performed in this area.

Experiment

Make the human subject ralk with arm swing restricted. A perturbation is applied to the trunk in order to imbalance the human. Recovery from perturbation is noted in the form of time required for achieving natural gait.

Repeat with the arm swing allowed.

8.2.4 Metabolic Energy Conservation with and Without AssistOn-Gait

Stroke patients have gait difficulties as well as impaired cardiorespiratory fitness. This may limit their excercise tolerance and may become a reason for limited effectiveness of RAGT (Robot assisted gait training). Restricting pelvic movements, as it happens in most of the RAGT devices, also results in more energy consumption.

Experiment

Make human subjects walk under Free Walking and Assisted Walking. Note energy consumption and develop an energy efficient gait. Analyse breathing (facemask, chest carrying breath-by-breath gas analysis system) and heart rate. No specialized hardware is required, commercially available sensors used by athletes can be used.

8.2.5 Evaluation of Effects of Inertia Compensation during Body Weight Support

Active body weight support systems provide a constant support despite vertical pelvic movements. This accounts for gravitational forces but not inertial forces. As human body moves vertically it acts as an inertial load. Ground reaction forces deviate due to inertia and must be compensated for. In Gravity-Assist this has been performed using emulated inertia compensation. Experiments have been performed and ground reaction forces have been estimated.

Experiment

Human subject walks with Active Body Weight Support. The experiment is repeated with Inertia Compensation switched On. Ground reaction forces are measured and compared to published data available on ground reaction forces. Commercial in-shoe force sensing systems are very costly. A low-cost alternative needs to be designed using thin capacitive force sensors.

Appendices

A Configuration and Motion Level Kinematics of 3$\underline{\text{R}}$RP Mechanism

Configuration and motion level kinematics of 3$\underline{\text{R}}$RP have been presented in [70, 71] in detail. Here, we summarize these kinematic analysis results for completeness. Figure A.1 depicts the kinematic schematics of 3$\underline{\text{R}}$RP planar parallel mechanism. 3$\underline{\text{R}}$RP mechanism consists of one base body, R, three body constituting the arms of the mechanism, Q, V, T and a symmetric end effector U. Arms Q, V and T have simple rotations with respect to base frame R with angles q_1, q_2 and q_3, respectively. These angles are actuated via motors that turns disks of the 3$\underline{\text{R}}$RP mechanism. Symmetric end-effector body U is connected to arm bodies from points Γ, Λ and Π via collocated linear and revolute joints. While point O is fixed on the base body R, S is the point at the middle of the end-effector body U of 3$\underline{\text{R}}$RP mechanism. End-effector body has transitions with respect to the base body about x_s at the direction of $\overrightarrow{r_1}$ and z_s at the direction of $\overrightarrow{r_3}$, also end-effector body U is rotated by θ around the axis of $\overrightarrow{r_2}$.

Fixed arm lengths of bodies between points of $O\Gamma$, $O\Pi$ and $O\Lambda$ are defined as l_1, l_2 and l_3. Variable distances between points ΓS, ΠS and ΛS are indicated as s_1, s_2 and s_3 respectively. In the kinematic calculation variable distances depicted above is assumed to be always positive as shown, while

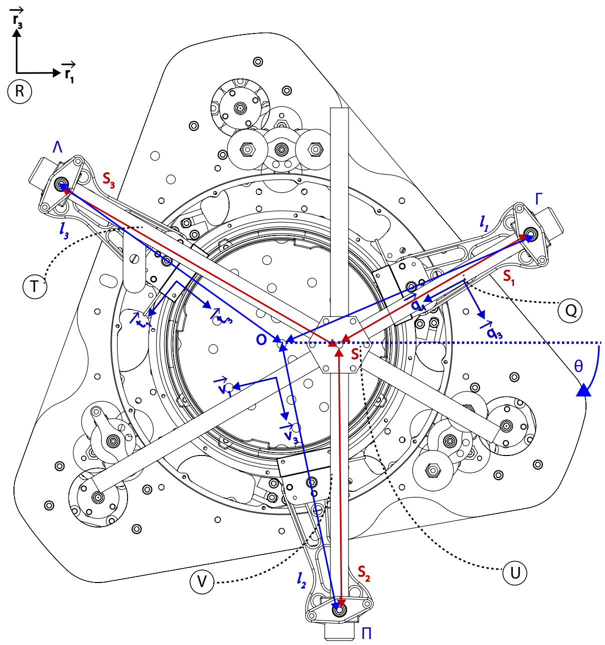

Figure A.1: Schematic representation of the kinematics of 3$\underline{\text{R}}$RP mechanism

angles are positive if counter-clockwise.

At the initial configuration, homing position, when $\overrightarrow{r_1}$ vector of base frame and $\overrightarrow{u_1}$ of end-effector body are overlapping with each other, angle θ is zero. Also at the homing position, the end-effector of 3$\underline{\text{R}}$RP mechanism starts from $x_s = 0$,$z_s = 0$, while arm vectors $\overrightarrow{q_1}$, $\overrightarrow{v_1}$ and $\overrightarrow{t_1}$ have angles $\pi/3$, π and $5\pi/3$ from base body vector $\overrightarrow{u_1}$.

A.1 Configuration level kinematics of 3$\underline{\text{R}}$RP mechanism

Forward kinematics at configuration level calculates end-effector configuration given input joint angles. According to [71], given arm angles q_1, q_2 and q_3, configuration level forward kinematics (end effector variables x_s, z_s and θ) can be analytically calculated as

$$x_s = -\frac{M}{\sqrt{3}(K^2 + L^2)} \tag{1}$$

$$z_s = c_{22} - \frac{K}{L}c_{21} - \frac{KM}{\sqrt{3}L(K^2 + L^2)} \tag{2}$$

$$\theta = atan2(K, L) \tag{3}$$

where

$$\begin{aligned}
K =& c_{12} + c_{32} + \sqrt{3}c_{31} - 2c_{22} - \sqrt{3}c_{11} \\
L =& c_{11} + c_{31} + \sqrt{3}c_{12} - 2c_{21} - \sqrt{3}c_{32} \\
M =& L(L - \sqrt{3}K)c_{12} - L(K + \sqrt{3}L)c_{11} \\
& - (L - \sqrt{3}K)(Lc_{22} - Kc_{21})
\end{aligned}$$

$$\begin{aligned}
c_{11} &= l_1 \cos(q_1) \qquad & c_{12} &= l_1 \sin(q_1) \\
c_{21} &= l_2 \cos(q_2) \qquad & c_{22} &= l_2 \sin(q_2) \\
c_{31} &= l_3 \cos(q_3) \qquad & c_{32} &= l_3 \sin(q_3)
\end{aligned}$$

After results of configuration level forward kinematic obtained, intermediate variables s_1, s_2 and s_3 can be calculated analytically using trigonometric relations.

Configuration level inverse kinematics calculates arm angles given the end-effector configuration of the mechanism. In particular, actuator angles

q_1, q_2 and q_3 can be found with given x_s, z_s and θ as

$$q_1 = atan2(M_1, L_1) \tag{4}$$

$$q_2 = atan2(M_2, L_2) \tag{5}$$

$$q_3 = atan2(M_3, L_3) \tag{6}$$

where

$$\begin{aligned}
K_1 &= x_s \sin(\theta + \frac{\pi}{3}) - z_s \cos(\theta + \frac{\pi}{3}) \\
K_2 &= x_s \sin(\theta + \pi) - z_s \cos(\theta + \pi) \\
K_3 &= x_s \sin(\theta - \frac{\pi}{3}) - z_s \cos(\theta - \frac{\pi}{3}) \\
M_1 &= K_1 \cos(\theta + \frac{\pi}{3}) - \sqrt{l_1^2 - K_1^2} \sin(\theta + \frac{\pi}{3}) \\
L_1 &= -K_1 \sin(\theta + \frac{\pi}{3}) - \sqrt{l_1^2 - K_1^2} \cos(\theta + \frac{\pi}{3}) \\
M_2 &= K_2 \cos(\theta + \pi) - \sqrt{(}l_2^2 - K_2^2) \sin(\theta + \pi) \\
L_2 &= -K_2 \sin(\theta + \pi) - \sqrt{(}l_2^2 - K_2^2) \cos(\theta + \pi) \\
M_3 &= K_3 \cos(\theta - \frac{\pi}{3}) - \sqrt{l_3^2 - K_3^2} \sin(\theta - \frac{\pi}{3}) \\
L_3 &= -K_3 \sin(\theta - \frac{\pi}{3}) - \sqrt{l_3^2 - K_3^2} \cos(\theta - \frac{\pi}{3})
\end{aligned}$$

A.2 Motion level kinematics of 3$\underline{\text{R}}$RP mechanism

Motion level kinematics is responsible for determining the linear relationship between actuator velocities and end-effector velocities. For the planar parallel mechanism, time derivative of configuration level kinematic equations can be utilized to solve for its motion level kinematics. In particular, the relationship between end-effector velocities $\dot{x_s}$, $\dot{z_s}$ and $\dot{\theta}$ and actuator velocities $\dot{q_1}$, $\dot{q_2}$, $\dot{q_2}$, as well as velocities of intermediate variables $\dot{s_1}$, $\dot{s_2}$, $\dot{s_2}$ for 3$\underline{\text{R}}$RP mechanism can be calculated as

$$\dot{X} = J_1^{-1} J_2 \dot{Q} \tag{7}$$

where

$$J_1 = \begin{bmatrix} 1 & 0 & -s_1 \sin(\theta + \frac{\pi}{3}) & \cos(\theta + \frac{\pi}{3}) & 0 & 0 \\ 0 & 1 & s_1 \cos(\theta + \frac{\pi}{3}) & \sin(\theta + \frac{\pi}{3}) & 0 & 0 \\ 1 & 0 & -s_2 \sin(\theta + \pi) & 0 & \cos(\theta + \pi) & 0 \\ 0 & 1 & s_2 \cos(\theta + \pi) & 0 & \sin(\theta + \pi) & 0 \\ 1 & 0 & -s_3 \sin(\theta - \frac{\pi}{3}) & 0 & 0 & \cos(\theta - \frac{\pi}{3}) \\ 0 & 1 & s_3 \cos(\theta - \frac{\pi}{3}) & 0 & 0 & \sin(\theta - \frac{\pi}{3}) \end{bmatrix} \tag{8}$$

and

$$J_2 = \begin{bmatrix} -l_1 \dot{q_1} \sin(q_1) & -l_2 \dot{q_2} \sin(q_2) & -l_3 \dot{q_3} \sin(q_3) \\ l_1 \dot{q_1} \cos(q_1) & l_2 \dot{q_2} \cos(q_2) & l_3 \dot{q_3} \cos(q_3) \end{bmatrix} \tag{9}$$

while

$$\dot{X} = \begin{bmatrix} \dot{x_s} & \dot{z_s} & \dot{\theta} & \dot{s}_1 & \dot{s}_2 & \dot{s}_3 \end{bmatrix}^T \text{ and } \dot{Q} = \begin{bmatrix} \dot{q}_1 & \dot{q}_2 & \dot{q}_3 \end{bmatrix}^T \tag{10}$$

The kinematic Jacobian is the matrix that maps joint velocities to end-effector velocities and frequently used for characterizing system and used in control algorithms. The kinematic Jacobian of 3$\underline{\text{R}}$RP mechanism can be found as

$$J_{3\underline{R}RP} = J_1^{-1} J_2 \tag{11}$$

At motion level inverse kinematics with given end-effector velocities, actuator velocities can be found. Motion level inverse kinematics is simple linear inverse of motion level forward kinematics; hence, it can be formulated as

$$\dot{Q} = J_{3\underline{R}RP}^{-1}\dot{X} \tag{12}$$

B Impedance and Admittance Type Devices

Impedance and admittance control are control strategies utilized for manipulating dynamic interaction between a system and the environment.

For impedance type systems the environment acts on a system and moves it, while the device reacts with an appropriate force. Therefore, the input to the system is displacement while the output is force. Impedance type devices are designed to have low weight, low friction and passive backdriveability (usually with cable driven transmission). They are very good at rendering low mass, but are not able to render high stiffness and forces [102].

Admittance type devices on the other hand operate in the opposite way. The environment exerts a force on them and they respond with motion. Therefore, force is input to the system and motion is the output. Using admittance control strategy and powerful actuators, these devices can be made large and heavy. Thus they are able to render large stiffness and forces, while offering a larger workspace than impedance type devices. They are not however passively backdriveable [102].

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