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Concept of a mobile robot-assisted gait rehabilitation system – simulation study

S. Slavnic, A. Leu, D. Ristic-Durrant, A. Gräser, Member IEEE

Abstract — **Neurological injuries caused by stroke, spinal cord injury or other illnesses and accidents often lead to walking disabilities. If not properly treated, gait disorders can lead to impaired physical and mental health, reduced physical activity, falls, fear of falling, loss of independence and the need for continuous medical care. After an accident or the start of an illness it is important to start the patient's gait rehabilitation process as soon as mental and physical conditions allow. Robotized rehabilitation systems have a number of potential benefits over traditional manual rehabilitation. These benefits include an increase in the effectiveness of the rehabilitation process, reduced costs of rehabilitation and reduced need for trained medical personnel.**

This paper presents details of the development of the concept of a mobile gait rehabilitation system consisting of a mobile base and powered exoskeleton. In particular, modeling and simulation of its mechanical system are considered. The presented simulation results demonstrate the need for including actuated translational degrees of freedom between the support platform and the exoskeleton to allow displacement of patient's pelvis and practice of dynamically balanced walking.

I. Introduction

THE recovery of walking skills after an accident or illness is usually done manually with the help of one or more therapists. The process is very long and exhausting for both patients and therapists. Unfortunately, it is also, very often, not completely successful for reasons such as: limited amount of walking possible in each therapy session, exhaustion, falling, injuries and the patient's fear of falling. In order to promote and make the process of gait rehabilitation more effective there has been a lot of effort, in recent years to develop robotic systems for gait rehabilitation.

The majority of the existing robotized gait rehabilitation systems that are commercially available (Lokomat [1], AutoAmbulator [2], GaitTrainer [3]) and those that are researched and developed but not commercialized (HapticWalker [4], Lopes [5], ALEX [6], PAM & POGO [7]) are stationary. Patients walk on a treadmill in exoskeleton-treadmill based systems [1], [2], [5], [6], [7] or legs are moved while attached to footplates in end-effector based systems [3], [4]. In each of these cases, the motion of patients between two points in space is not possible. In order to overcome this disadvantage, virtual reality (VR) systems have been applied for presenting different tasks to be executed by patients, which encourages and motivates patients [8].

In order to provide patients with the ability to move forwards during rehabilitation, a significant characteristic of normal walking, robotized mobile gait rehabilitation systems have been researched and developed. Different design concepts have been proposed including different combinations of a mobile support platform (base) with or without a passive or active exoskeleton. Some of these will be reviewed in this paper.

Mobile gait rehabilitation systems that consist of mobile platform without passive or active exoskeletons can be found in [12] and.[13]. These systems preserve the balance of the patient and are equipped with different mechanisms to help the patient during rehabilitation, such as weight reduction mechanism, motion intention detection, but they have no mechanism explicitly designed to help the patient to complete initiated leg movements, if the patient is not able to do that by themselves. This means that, the therapist must still move the patient's legs.

Active exoskeletons without a support base are rarely used for gait rehabilitation, although such systems have been researched with that aim [9]. They are instead usually used for human strength enhancement [10], [11]. The main disadvantages of independently using active exoskeletons in gait rehabilitation are the difficulty of preserving the patient's dynamical balance, and the patient's fear of falling.

Mobile gait rehabilitations systems that combine a mobile base with a powered exoskeleton have been rarely considered even though this combination would provide patients with both mobility and a mechanism to help them to complete intended leg movements. In this paper such a combination is considered. The results of simulations, the first critical step in demonstrating the feasibility of the system, are presented. The description focuses on the actuated degrees of freedom necessary to fulfill the requirements. Systems that support the patient weight and prevent patients from falling such as those described in [14] are not discussed here.

Different robot assisted gate rehabilitation systems provide the patient with different Degrees of Freedom (DoF)

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S. Slavnic, A. Leu, D. Ristic-Durrant, A. Gräser are with the Institute of Automation, University of Bremen, NW1, Otto-Hahn-Allee 1, 28359 Bremen, Germany (phone: +49-421-218-62474 e-mail: {slavnic, leu, ristic, ag}@ iat.uni-bremen.de).

during walking. Some rehabilitation systems equipped with active exoskeletons only have revolute joints of hip and knee actuated in sagittal plane [1]. Others have, in addition, the hip actuated as a revolute DoF in the frontal plane [6], [7]. Exoskeletons for rehabilitation are very often constructed without actuated foot segments [1], [5], [7], [9] and so the patient's ankle joints are not controllable. Normally, either a patient moves their foot freely if they are capable of controlling their ankle joints, or the foot is lifted using passive lifting mechanisms (elastic straps or passive orthosis) during the swing phase of walking. In some systems motion of the pelvis is only allowed in the vertical direction (translation) [1] or in vertical and horizontal directions (translation) [5]. In contrast some systems provide many more DoF for the pelvis segment, such as [7], where all six DoF are allowed (three translations and three rotations). The greater the number of allowed DoF in the pelvis segment the more natural the walking motion can be, but, the control of a large number of DoF is complex and the required construction is bulky. A detailed overview of active orthosis for the lower limbs can be found in [15].

The reference values for the controllers used to guide the patient's legs are usually derived mathematically, according to the kinematic parameters of the patient. During this process it is assumed that the patient should move his/her legs similarly to the gait pattern of a healthy subject.

Hence, in existing gait rehabilitation systems, patients are forced to practice "stationary walking" on a treadmill, to move their legs similarly to the gait pattern of healthy subjects, and patients are not able to determine for themselves the gait pattern that best fits their particular physical and mental condition, and thus patients cannot practice dynamic balanced walking patterns that match their requirements. These are some of disadvantages of the existing gait rehabilitation systems that we aim to overcome by introducing a novel mobile gait rehabilitation system concept.

The aim of our feasibility study presented here is to develop the concept of a mobile gait rehabilitation system which would overcome the disadvantages of existing gait rehabilitation systems mentioned above. The system to be developed and researched should fulfill the following requirements:

- provision of body balance for the patient;
- provision of walking assistance through a powered exoskeleton;
- provision of a body weight reduction mechanism;

Furthermore, the system should:

- be applicable to a wider range of gait disorders;
- allow patient to select their required gait pattern;
- allow the practice of dynamic balanced walking.

The novelty in our approach is the consideration of practicing dynamic balanced walking during rehabilitation. The final aim of our research is to control the patient's balance during walking if they are not able to do that by themselves. However, in order to develop a system to fulfill the requirements above, it is necessary to make a compromise between medical (rehabilitation) requirements and technological plausibility.

II. Mechanical Design of the Rehabilitation System

The CAD mechanical design of the mobile gait rehabilitation system used in the simulation study is shown in Fig. 1. As shown, the system consists of a mobile platform with wheels and a powered exoskeleton. The rear wheels are drive wheels while the front wheels are used for steering. The exoskeleton consists of thigh, shank and pelvis segments that are interconnected by revolute joints. As a starting point for simulations, the number of actuated DoF was chosen to be the same as in [1]. This number of DoF enables realization of "human like" walking patterns while a lower number of allowed DoF would be inappropriate and inadequate for this. Thus, the actuated joints on the exoskeleton are revolute joints of the hip and knee in sagittal plane, and one actuated translational DoF exists in the connection between the exoskeleton and the mobile platform that allows vertical translation of the human body. Motions around the remaining rotational axes (frontal and transverse) of the exoskeleton's hip and knee joints are restricted by the mechanical construction of the exoskeleton.

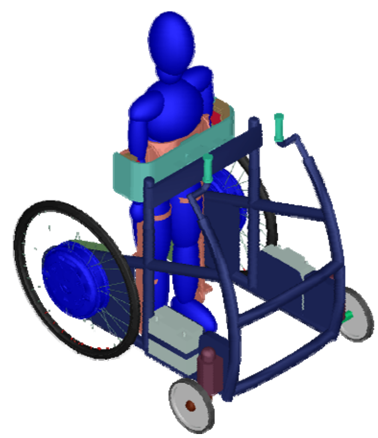

Fig. 1. Mechanical CAD design of the mobile gait rehabilitation system

III. Mechanical Models and Simulation Setup

A combination of MSC Adams [16], Lifemode [17] and MATLAB [18] software are used for modeling, simulation and optimization of the designed mechanical system. Lifemode software was used for creating human body models, foot-ground contacts and as an interface for importing human motion data. MSC ADAMS was used for simulating mechanical models of the rehabilitation system and the human body. Finally, MATLAB was used for pre-processing motion data and simulation results. This selection of simulation tools was found to be optimal for modeling all the components of the system and for observing the mechanical interaction between the system and the human body model. The development model is realistic, as all important effects are modeled including ground-foot contacts, ground-tire contacts, human body–exoskeleton contacts, joint boundary contacts and friction in joints

The mechanical CAD construction of the mobile platform and the exoskeleton are imported into MSC Adams to provide the geometrical and mass properties of the components. The ground-tire contacts are modeled using the MSC Tire module.

The mechanical model of the human body was initially generated using the LifeMode software database and has later been refined to suit the particular subject, for whom motion data are captured and whose walking inside the system is simulated. The mechanical model of the human body consists of 19 segments (links) that are interconnected with spherical joints. The spherical joints are modeled as a section of three revolute joints, between which are inserted "virtual" segments with negligible mass and inertial properties, so they do not affect the dynamics of the complete mechanism. For the knee joints only, two rotations around transverse and frontal axis are locked and the joint has one DoF around the sagittal axis. The modeling of the ground-foot contacts is done using 11 contact points per foot segment equipped with contact ellipsoids. Normal and transverse friction forces are generated according to the depth of penetration of the ellipsoids into the ground surface.

The mechanical models of the human body and the system are interconnected using connection elements with stiffness-damping characteristics. The stiffness and damping coefficients of these elastic elements are set such that only a small displacement between the human segments and segments of the exoskeleton exist. In this way, the human body model is coupled to the mechanical model of the rehabilitation system through elastic elements rather than through rigid connection elements. At every connection point on the human body model and the rehabilitation system model coordinate systems are placed, and distances and rotations between these coordinate systems are measured and action-reaction forces are generated that act on the exoskeleton and human body according to following equations:

$$\mathbf{F_c} = \mathbf{k} \cdot \mathbf{d} + \mathbf{c} \cdot \dot{\mathbf{d}} \quad (1)$$

$$\mathbf{T_c} = \mathbf{k} \cdot \mathbf{r} + \mathbf{c} \cdot \dot{\mathbf{r}} \quad (2)$$

where, $\mathbf{F_C}$ and $\mathbf{T_C}$ are action-reaction forces and moments respectively, $\mathbf{k}$ and $\mathbf{c}$ are the stiffness and damping coefficients (500N/m, 50Ns/m), $\mathbf{d}$ is the distance vector between contact points on human body and exoskeleton and $\mathbf{r}$ is the vector containing rotation angles between local coordinate systems that reside in contact points. Two of these connection points exist for each leg segment and several of them exist on the pelvis segment. The stiffness and damping coefficients are tuned by running the simulation several times and observing the simulation results.

The motion capture data used in simulations were obtained by a markerless motion capture system developed at our institute [19], which can reliably extract the gait features from frontal and sagittal images of subjects walking in a natural environment. The extracted gait parameters were the joint angle changes of the hip, knee and ankle.

IV. PRACTICING DYNAMICALLY BALANCED WALKING INSIDE THE REHABILITATION SYSTEM

In order to allow the patient to practice walking inside the rehabilitation system in a way that is also feasible outside the system (walking without falling), additional actuated degrees of freedom are necessary, besides those initially selected, as will be explained in this section. Human walking is dynamically balanced as long as the zero moment point (ZMP) resides inside the support area [20]. The support polygon is a convex hull of all contact points between the feet (a foot in the single support phase) and the ground. The ZMP is the point inside the support area at which the net moment due to gravitation and all inertial forces induced by the mechanism's motion have no components along the horizontal axes, and the support area is the surface determined by the contact of the foot and ground. Fig. 2 shows a sketch of a human in the single support phase of a gait cycle (when only one leg is in contact with the ground) and the forces and moments that act on human body during walking. The human body is modeled as a rigid-body system.

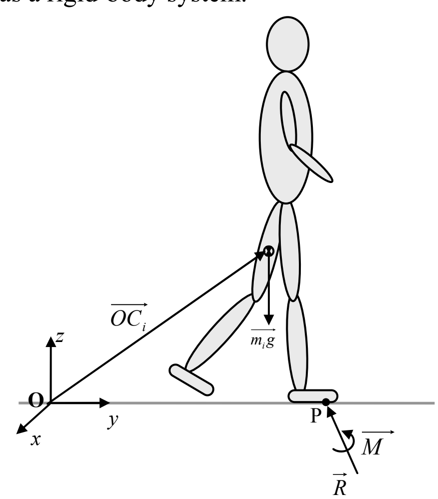

Fig. 2. Single support phase: human body and forces and moments acting on it

The Vector $\overrightarrow{OP}$ is the position vector where the resultant ground reaction force $\mathbf{R}(R_x, R_y, R_z)$ and the moment $\mathbf{M}(M_x, M_y, M_z)$ are acting on the foot segment. The vector $\overrightarrow{OC_i}$ is the position vector of the center of masses (CoM) of the *i*-th segment. All vectors are expressed in the global coordinate system **O**.

$$\overrightarrow{OP} = [x_p, y_p, z_p] \quad (3)$$

$$\overrightarrow{OC_i} = [x_i, y_i, z_i] \quad (4)$$

The following equations describe the dynamic equilibrium of the system during motion in the global reference system:

$$\vec{R} + \sum_i (\vec{F_i} + \vec{G_i}) = 0 \quad (5)$$

$$\overrightarrow{OP} \times \vec{R} + \sum_i (OC_i \times (\vec{F_i} + \vec{G_i})) + \sum_i \overrightarrow{M_i} + \vec{M} = 0 \quad (6)$$

Vectors $\vec{F_i}$ and $\overrightarrow{M_i}$ represent the inertial force and moment of inertial force acting on *i*-th segment while $\vec{G_i}$ is the gravitational force of the *i*-th segment acting at the center of mass. Taking $\overrightarrow{M_x}$ and $\overrightarrow{M_y}$ equal to zero, the coordinates of the ZMP on the ground can be found using the following two equations:

$$x_p = \frac{\sum_i (m_i x_i (\ddot{z}_i + g) - m_i z_i \ddot{x}_i) - \sum_i M_{iy}}{\sum_i m_i (\ddot{z}_i + g)} \quad (7)$$

$$y_p = \frac{\sum_i (m_i y_i (\ddot{z}_i + g) - m_i z_i \ddot{y}_i) + \sum_i M_{ix}}{\sum_i m_i (\ddot{z}_i + g)} \quad (8)$$

Fig. 3 shows simulation results for the ZMP position and the projection of the body's CoM on the ground plane during human walking at approximately 4 km/h. As can be seen, the ZMP goes from one support leg to the other (illustrated with a rectangular box in Fig. 3) while the projection of the CoM shows only small displacements in the plane normal to the walking direction.

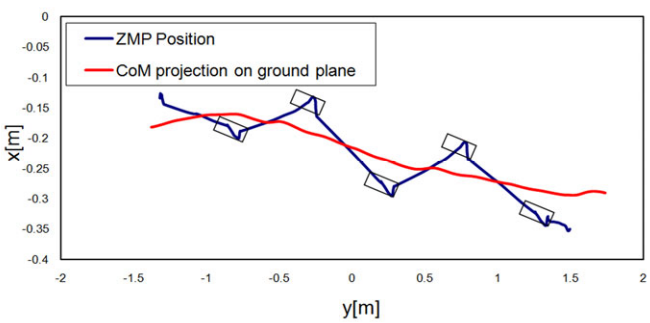

Fig. 3. ZMP position and CoM projection on ground plane for model of human walking at ~4km/h

From equations (7) and (8) stands that in the stationary state ZMP is at the same point as the projection of the CoM on the ground since all acceleration components are equal to zero. Also, it can be concluded that for lower walking velocities the terms that depend on accelerations in equations (7) and (8) will tend to zero. Because the ZMP for dynamical balanced walking has to stay inside the support polygon, the CoM must be displaced by a higher magnitude horizontally in the frontal plane. This coincides with measurements of CoM displacement during walking at different walking speeds [21]. The CoM displacement in the case of irregular walking (that is, the subject walks with some form of walking disability) can be much larger in magnitude than is the case for healthy subjects, as illustrated in Fig. 4. The Fig 4 shows the CoM displacement of a subject who walks with an irregular gait pattern due to cerebral palsy (CP) compared to a healthy subject [19]. The subjects walking is dynamically balanced even though, the gait pattern is irregular. Our idea and intention is to allow patients to practice irregular gait patterns in the rehabilitation system as long the walking is dynamically balanced. In other words, if as a result of physical constraints the patient can not achieve walking with a regular gait pattern, but can nonetheless walk in a dynamically balanced way with an irregular gait pattern, then the rehabilitation system should allow patient to practice the gait pattern which best suits their physical state, as this can be more beneficial to the patient.

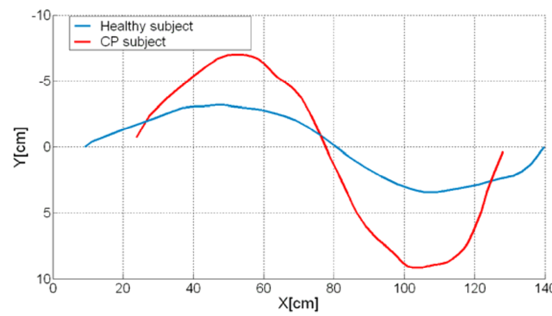

Fig. 4. CoM displacement of subjects that walk irregularly (CP) and healthy subject

From the above discussion it follows that, in order to develop a robotic gait rehabilitation system that allows patients to practice dynamically balanced walking, it is necessary to include an additional translational DoF between the exoskeleton and the mobile platform that will allow the patient to move their pelvis and upper body horizontally in the frontal plane. Also, an additional revolute DoF in the frontal plane has to be added to the hip joints so that the feet do not lose contact with ground when the pelvis translates in the frontal plane.

V. SIMULATION RESULTS

A. First Simulation scenario

The first simulation scenario was walking inside the rehabilitation system when the only allowed DoF were the vertical translation of the pelvis, and revolute motions of the hip and knee in the sagittal plane.

Motion data of a healthy subject (without physical restrictions) have been found to be unsuitable for simulating walking inside the system, because the human body model in the system can use only limited number of DoF during walking, compared to healthy subjects that use all available DoF. Therefore, the human subject (178cm height, 91kg weight), for whom walking inside the system has been modeled, imitated walking inside the rehabilitation system by limiting motions in all those joints in which motion is restricted by the mechanical construction of the system. Motion data for ankle, knee and hip joints were extracted from the frontal and sagittal plane. The extracted motion data were used to kinematically drive the model's hip and knee joints in the sagittal plane and the ankle joints in the frontal and sagittal planes. Fig. 5 depicts the angle changes of the right leg's hip and knee used for simulation, that are extracted using the markerless motion capture system from a movie of the subject walking. The transitional motion of the pelvis was calculated according to the lengths of the left and right legs during walking, such that at least one foot is always in contact with ground.

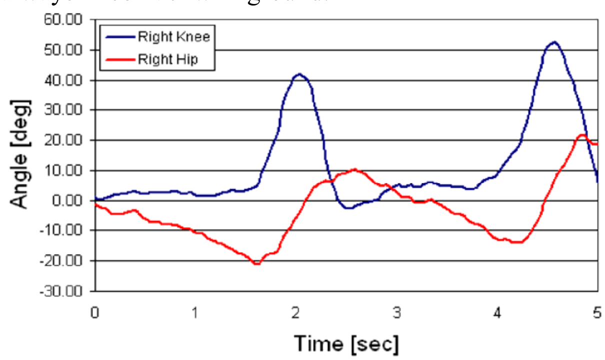

Fig. 5. Angle changes in hip and knee used for simulation scenario 1

Fig 6 shows the simulated ground reaction force (GRF) during walking inside the system for the first simulation scenario. At some moments of the swing phase both feet come in contact with ground because the CoM translation in the system is restricted, and during walking the humans translate their CoM horizontally in the frontal plane.

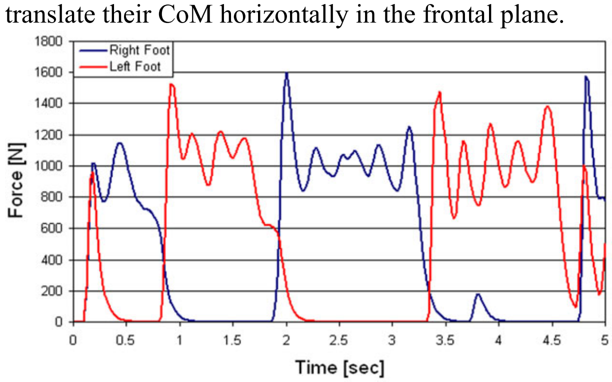

Fig. 6. Ground reaction forces during walking inside the rehabilitation system obtained by simulation in scenario 1

B. *Second simulation scenario*

In this simulation scenario the horizontal translation of the pelvis and rotational motions of the hip joints in the frontal plane are allowed, besides the DoF used in the first simulation scenario. In this case the motion data of a healthy subject (137cm height, 36kg weight) walking "normally" at moderate speed are used. Fig. 7 shows the angle changes in the hip and knee joints used for simulation. The horizontal translational motion of the pelvis is mathematically generated, and its amplitude is larger than it would be in a case of "normal" walking at simulated speed, in order to emphasize the functionality of the translational DoF. As in the previous simulation scenario, motion data are used to drive the kinematically allowed DoF.

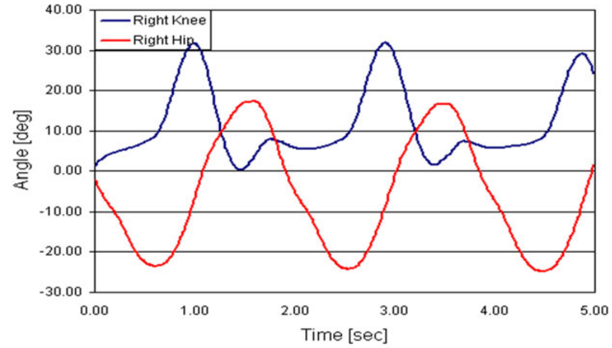

Fig.7. Angle changes in hip and knee used for simulation scenario 2

Fig. 8 shows the simulated resultant ground reaction force during walking inside the rehabilitation system. The figure shows that both legs have proper contact with ground during walking. The shape of the GRF shows the first "impact" peak while the second peak was omitted, comparing to the regular walking [22]. This is caused by imprecise motion data that are used for driving the mobile platform and the pelvis translational joint. The contact of the swing leg with the ground does not occur anymore because of the additional DoF available.

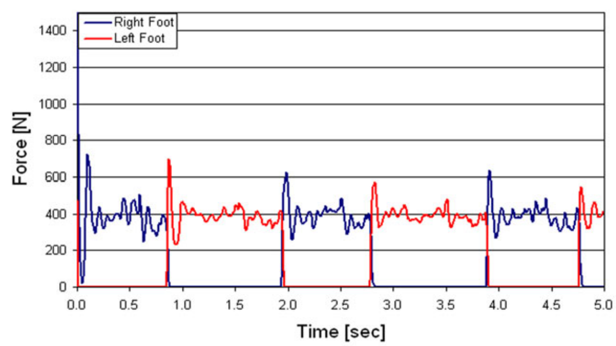

Fig. 8. Ground reaction forces during walking inside the rehabilitation system obtained by simulation in scenario 2

Fig 9 shows the horizontal displacements of the subject's CoM in the frontal plane for both the simulation scenarios considered.

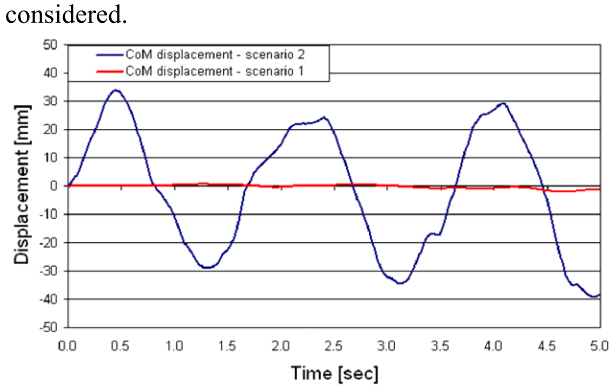

Fig. 9. CoM displacement for simulation scenario 1 and 2

The small displacement of the CoM in simulation scenario 1 is a consequence of the elasticity of stiffness-damping elements inserted between the human body and the exoskeleton that permits only small movements. The CoM displacement in simulation scenario 2 is as expected large due to the additional DoF.

VI. Conclusions and Future Work

In this paper the concept of a novel mobile gait rehabilitation system has been proposed and simulation results of a human walking inside the proposed rehabilitation system have been presented. The mobile gait rehabilitation system consists of a mobile platform and an exoskeleton. Simulations have been performed for different sets of allowed degrees of freedom, namely the degrees of freedom that can be used by the human model while walking subject to the constraints of the rehabilitation system's mechanical construction. The results show that in a case when the horizontal translation of the pelvis is not allowed, the human model is able to walk inside the system, and the support platform provides balance during walking. However, walking with the same gait pattern would not be feasible without the support platform because the patient would fall due to the suppressed pelvic horizontal movement. Humans displace their CoM during walking and by controlling CoM displacement they preserve dynamic balance. Therefore, an additional translational degree of freedom is added that allows the pelvis segment to move horizontally in the frontal plane. In addition, revolute joints are added to the hips in the frontal plane to preserve the full surface contact of the feet with the ground when the pelvis translates horizontally. Additional degrees of freedom allow the practice of walking inside the system that also could be performed without the help of the system. In this way patients would be able to practice "balancing" inside the system that could result in more effective rehabilitation and better and safer walking after finishing the rehabilitation process.

Our future work will be concentrated initially on the modeling of the actuation system (which will be consisting of electromechanical actuators), estimation of patient balance and the development of control systems for the rehabilitation device. Different controllers will be developed for different rehabilitation stages. The platform will be actuated depending on the patient intention to walk and a given task, such that the mobile platform will not restrict the patient's motion. For this purpose a force/torque sensor will be inserted at the connection point between the exoskeleton's pelvis segment and the mobile platform. All these subsystems will be simulated and proved in the presented simulation environment.

VII. Acknowledgment

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