LOPES II—Design and Evaluation of an Admittance Controlled Gait Training Robot With Shadow-Leg Approach

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***Abstract*—Robotic gait training is gaining ground in rehabilitation. Room for improvement lies in reducing donning and doffing time, making training more task specific and facilitating active balance control, and by allowing movement in more degrees of freedom. Our goal was to design and evaluate a robot that incorporates these improvements. LOPES II uses an end-effector approach with parallel actuation and a minimum amount of clamps. LOPES II has eight powered degrees of freedom (hip flexion/extension, hip abduction/adduction, knee flexion/extension, pelvis forward/aft and pelvis mediolateral). All other degrees of freedom can be left free and pelvis frontal- and transversal rotation can be constrained. Furthermore arm swing is unhindered. The end-effector approach eliminates the need for exact alignment, which results in a donning time of 10–14 min for first-time training and 5–8 min for recurring training. LOPES II is admittance controlled, which allows for the control over the complete spectrum from low to high impedance. When the powered degrees of freedom are set to minimal impedance, walking in the device resembles free walking, which is an important requisite to allow task-specific training. We demonstrated that LOPES II can provide sufficient support to let severely affected patients walk and that we can provide selective support to impaired aspects of gait of mildly affected patients.**

***Index Terms*—Admittance control, gait training, haptics, robotics, spinal cord injury, stroke, zero impedance.**

I. Introduction

Since the end of the 20th century gait training robots have been developed [1], [2]. These devices reduce the physical workload for the therapist. Studies on effectiveness have shown contradictory results [3]–[7]. Recent meta-analyses have shown that for spinal chord injury (SCI) patients robotic-aided gait training has no beneficial effect compared to conventional therapy [8], but for stroke survivors robotic gait training is beneficial especially in the sub-acute phase and for severely impaired patients [6], [9].

To improve the efficacy, gait training robots should encourage the patient to actively participate [10], [11]. This can be achieved by an "assist-as-needed" (AAN) approach [12]–[15]. For severely affected patients, this implies that the robot should provide much assistance. This can be achieved with feedforward force [16], [17] or with high stiffness (high impedance). For mildly affected patients, the robot should behave transparently (low impedance) and provide assistance only on aspects that require support. Robot mechanics and control are key-drivers in facilitating AAN [18].

Implementation of AAN requires the robot mechanics to allow free motions in all degrees of freedom (DoFs) of gait, and when needed, to provide support in most DoFs. Hindering free motions or constraining DoFs causes changes in gait kinematics [4], [19]. Abduction and pelvis translations are constrained in e.g., the first generation Lokomat [1] and the Reo Ambulator [20]; Pelvis rotations are constrained in most devices e.g., Lopes I [21], Lokomat and Reo Ambulator. ALEX allows for pelvis vertical rotation [22], but not for pelvis anterior/posterior translation and other rotations. The PAM and POGO allows for free motion in all DoFs [17]. *Free motions* also means that arm swing should be possible, since it is part of normal walking and it contributes to the overall stability of human gait [23]. In most exoskeleton gait training robots arm swing is obstructed by the presence of mechanics beside the hip joints [4].

The interaction forces between robot and patient should be force controlled to implement both low and high impedance control [11], [24], [25]. The Lokomat was originally position controlled (high impedance), later force control was applied to the Lokomat [26]. This reduced the impedance of the Lokomat, however the behavior is not sufficiently transparent [10]. Several devices have been designed with low-impedance control as starting point, however, they all compromise on high impedance support: The stiffness of the Series Elastic Actuation in Lopes I is limited [21], [27], and therefore the high impedance mode is not stiff enough for training severely impaired patients. Similarly the PAM and POGO perform well in the low impedance control [17], but the stiffnesses are limited [28] and insufficient for high impedance control in gait training. Sulzer *et al.* [29]

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focused on low impedance for an active knee support, compromising high impedance support. AAN has been tested successfully on stroke-survivors with mild to moderate impairment with the single-sided exoskeleton ALEX [30]. The challenge is to develop a gait training robot that is both sufficiently transparent and sufficiently stiff. Little has been published on the required transparency and stiffness. Regarding the required transparency, we have previously assessed the maximum allowable inertia added to the pelvis and ankle that do not give significant changes in kinematics during walking at 4.5 km/h [31].

Another aspect that needs improvement to optimize robot-aided gait training is the donning and doffing time. To increase the usability of gait training robots in practice, the donning time should be reduced. Little has been published about donning time of gait training robots. Nilsson *et al.* [32] reported a donning time of 15–20 min for the HAL robot, but this includes application of EMG. The donning time of the PAM and POGO is up to 30 min [17]. In interviews users of the Lokomat and Lopes I reported donning times of 20 min for the first training (including limb measurement and cuff selection) and 10–15 min for recurring trainings. These donning times are considerable given the duration of training sessions (30–60 min). In exoskeleton robots the long doffing/donning times are caused by the need to precisely align the robot joint axes with the human joints to prevent damage and uncomfortable man-machine interaction [33]. Improving the donning time is likely to be less enduring for both patient and physiotherapist.

Our goal was to design and evaluate a gait training robot suitable for clinical training of severely and mildly affected patients. This implies that donning should be quick, multiple DoFs should be free and the controller must be able to switch between low impedance (no support) and high impedance (full support).

The paper is structured as follows. In Section II we discuss the requirements for the gait training robot. Section III describes the mechatronic design of LOPES II. Section IV describes the controller. Section V describes the validation of the technical requirements and the clinical feasibility of LOPES II. Section VI is the discussion. Finally conclusions are drawn in Section VII.

II. Requirements

We visited three rehabilitation centers and interviewed several physiotherapists, rehabilitation physicians, stroke survivors and researchers to obtain the user requirements. The user requirements were gathered with observations, interviews and group sessions with physiotherapists, stroke survivors, researchers and rehabilitation physicians. The main user requirements are 1) quick donning and doffing; 2) allow for motions occurring in a variability of (pathological) gait (e.g., hip-hiking and circumduction); 3) provide assist-as-needed for patients with functional ambulation category (FAC) 0–5; and 4) ensure safety. The requirements are quantified in the following sections.

A. Donning Time

According to therapists and rehabilitation physicians, the time needed from the start of the preparation to the start of the training (donning time) should be less than 10 min when patient come in for the first time and less than 5 min for recurring visits. The first time takes more time as it includes measurements of segment lengths and adjustments of the device.

TABLE I
Requirements Per Degree of Freedom in Terms of Range of Motion, Torque and Speed (Powered DoFs Only)

DoF	RoM [deg; m]	Force/ Torque [N; Nm]	Speed [rad/s; m/s]
Pelvis anterior/posterior	±0.2	500	0.3
Pelvis mediolateral	±0.15	500	0.3
Pelvis up / down	±0.1	1000[a]	
Pelvis sagittal rotation	±6		
Pelvis frontal rotation	±10		
Pelvis transversal rotation	±15		
Hip abduction / adduction	20 / 20	70	
Hip flexion / extension	40 / 30	70	3.2
Hip endorotation / exorotation	15 / 15		
Knee extension / flexion	0 / 75	70	7.3
Foot endorotation / exorotation	10 / 20		
Ankle dorsiflexion/ plantarflexion	25 / 35		
Ankle inversion / eversion	10 / 10		

[a]Upward only (body weight support)

B. DoF Requirements

Table I lists the requirements per DoF. For the joint excursion limits we used the 95% interval of data from normal walking [34], [35] and from paretic gait [36]–[38]. For the pelvis translations we increased the range of motion to allow for acceleration/deceleration and drifting sideways.

Transparent behavior of the robot implies that the patient should be able to move freely with minimal resistance (impedance) of the robot [24]. With control strategies the robot impedance can be compensated for largely, but not completely. The remaining impedance can be implemented as an inertia, a damper or a combination of both. If the remaining impedance is sufficiently low, the gait pattern will not notably be affected. Little has been published on the maximum allowable impedances for gait training; we only have data about impedance expressed in terms of inertia: for pelvis anterior posterior and lateral motions is 6 kg and for the foot anterior translation it is 2 kg [31]. For the knee it is unspecified, but since the accelerations of the knee are less than that of the foot, but more than that of the hip it is estimated at 4 kg.

C. Anthropometric Data

To cover a wide population, we used several anthropometric datasets [39]. We aimed to cover 99% ($\mu \pm 3\sigma$) of the populations of Europe and North America (see Table II).

III. Design Description

In order to minimize the moving mass of the robot, we incorporated fixed base actuation in LOPES II. We used a structure of push-pull rods, to transfer the motor torques to the patient, since they are light-weight and stiff. The shadow leg [41] serves as an intermediate body between motors and patient (see Fig. 1). Each motor is connected to a single segment of the shadow leg. The segments of the shadow leg are connected to the segments of the

TABLE II
ANTHROPOMETRIC DATA COVERING 99% OF THE EUROPEAN AND NORTH AMERICAN POPULATION

	Min	Max
Stature [mm]	1410 [a]	2088 [b]
Mass [kg]	36 [c]	138 [d]
Shank Length [mm]	347[e]	514[e]
Thigh Length [mm]	345[e]	512[e]

[a]Dutch 2004 (60 plus), female
[b]Dutch 2004 (20-30 years), male
[c]Dutch 2003 (31-65 years), male
[d]Dutch 2003 (31-65 years), male
[e]Values derived from stature [40]

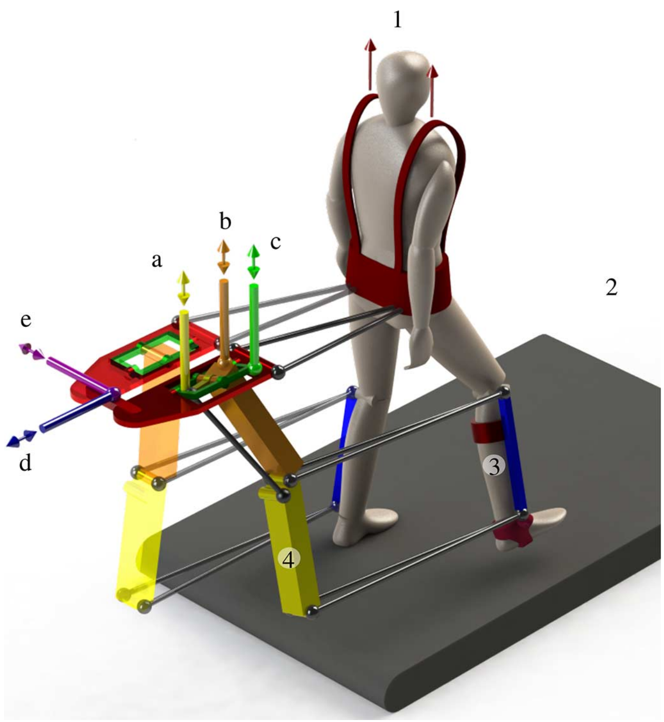

Fig. 1. Schematic overview of LOPES II with the shadow leg. Patient is placed on a treadmill (2) and attached to the harness with body weight support (1). Patient is clamped below the knees and at the feet. Clamps are connected to a leg guidance (3), which is connected to the shadow leg (4) with horizontal push pull rods. Shadow leg is actuated in shank flexion/extension (a), thigh flexion/extension (b) and abduction/adduction (c). Shadow leg is suspended on a stage connected to the patient's pelvis with rods, actuated in pelvis forward/aft- (d) and mediolateral direction (e).

patient leg (Fig. 1). The angles of the shadow leg are directly related to the angles of the patient leg, and thus given the patient limb lengths and angles of the shadow leg, the patient joint angles can be calculated exactly [42].

The shadow leg offers several advantages. The first is that a minimal amount of clamps are needed to control the patient's leg. We chose to have clamps at the feet, at the lower legs just below the knees, and at the pelvis (see Fig. 1). The second advantage is that, due to the minimum amount of clamps and the location of the shadow leg behind the patient, a small misalignment of the rods (i.e., not pointing exactly to the patient joints) will not cause any strains in the patient. Also these misalignments have only little effect on the calculated joint angles, e.g., on a lower leg of 400 mm, a vertical misalignment of 10 mm on the knee joint gives a 2.5% error in the calculated shank and thigh angle. Since all mechanics are located behind the patient, the arm swing is unhindered. The last advantage is that a physical end stop can be incorporated in the shadow leg to prevent hyper-extension of the knee.

TABLE III
ACTUATION OF THE LOPES II

DoF	Actuator	Gearing	Force / Torque at segment[a]
Pelvis AP	C100	0.4 m	250 N
Pelvis ML	C40	0.2 m	200 N
Abduction	C40	2/3	60 Nm
Thigh flexion	C40	2/3	60 Nm
Shank flexion	C100	3/2	66 Nm

[a]force torque values are continuous. The motor and drive are capable of delivering a peak torque that is double the continuous torque, for a short period ($< 1sec$).

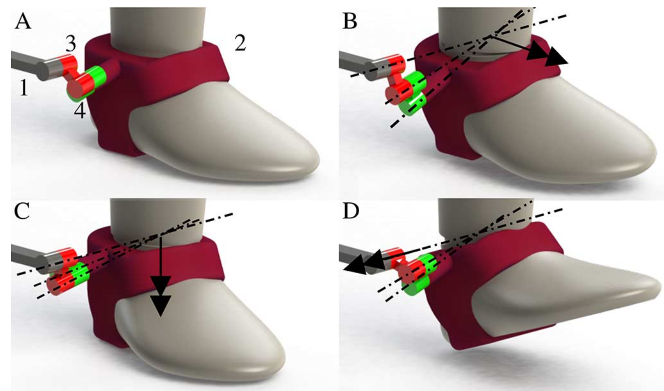

Fig. 2. Spherical gimbal connection of the ankle joint. Rod from the shadow leg (1) is connected to the foot bracket (2) with two segments (3, 4). Connections between the components are revolute joints, with axes intersecting in the ankle joint. This allows for rotation of the foot about the three principal axes: inversion/eversion (B), foot endorotation/exorotation (C) and plantarflexion/dorsiflexion (D).

For actuation we used Moog Control Loading actuators CL-R-E-/MD/40Nm (C40) for thigh flexion and abduction and pelvis lateral motion and CL-R-E/MD/100Nm (C100) for the shank flexion and the pelvis anterior posterior motion [43]. We used a gearing with rods between the actuators and the human segments (see Table III).

The actuators are linked to the segments as follows. The shadow hip is attached to a stage that is actuated in the horizontal plane (d and e in Fig. 1), this way pelvis horizontal forces are applied to the patient. Thigh flexion of the shadow leg (actuated by b in Fig. 1) pushes the shadow knee and thus the patient knee forward. Shank rotation of the shadow leg (actuated by a in Fig. 1) translates the shadow ankle and thus moves the patient ankle forward and backward. By applying abduction to the shadow leg (c in Fig. 1), the shadow ankle moves outward, and thus patient foot is pulled outward.

The rods pointing towards the ankle are connected to a foot bracket with a spherical gimbal, which has the center of rotation in the ankle joint. This assures that force from the ankle rods are exerted in the center of the ankle without imposing torques on the foot (see Fig. 2). At the pelvis a similar mechanism has

(a) (b) (c)

Fig. 3. LOPES II installed at rehabilitation center Roessingh, Enschede, The Netherlands (a). Patient is connected to LOPES II with a harness, clamps just below the knees, and foot brackets (b). Force sensors (1) are located closely to the patient, and the spherical gimbal at the pelvis (2) allows for rotations of the pelvis (c).

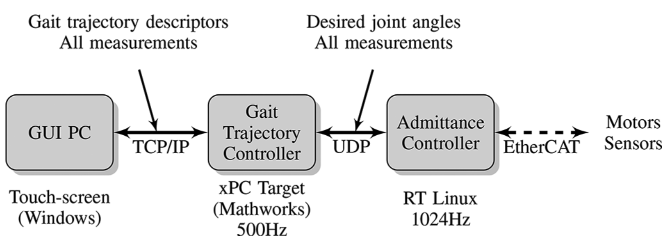

Fig. 4. Three computers used for controlling LOPES II.

been implemented to apply forces on the hip joints and allow for pelvic rotations in all three directions. At the end of the rods, near the patient, force sensors are mounted to measure the force between the patient and LOPES II (see Fig. 3).

The patient is suspended from a harness [(1) in Fig. 1], connected to pneumatic bodyweight support system (BWS). A near-constant upward force is applied to the harness by controlling the air-pressure in the BWS. Furthermore, the BWS contains a settable end-stop that limits the downward displacement of the harness to prevent falling.

Fig. 3 shows LOPES II as installed the Roessingh rehabilitation center in Enschede, the Netherlands. A second system has been installed in the Sint Maartenskliniek in Nijmegen, the Netherlands.

IV. Controller Description

LOPES II is controlled by three interconnected computers, as shown in Fig. 4. First, the graphical user interface is a touch screen PC with which the operator can adjust the level of required support for each individual patient. Additionally it provides real-time feedback on the patient's performance and allows for post analysis of recorded data. Second, the gait trajectory controller generates patient-specific gait patterns and support patterns. Third, the admittance controller converts the gait and support patterns to actuator set points and processes data from the sensors. In the following sections, the three computers are discussed in detail.

TABLE IV
List of Available Sliders for Adjusting the Gait Trajectory and Support

Gait subtask	Support Adjustment	Parameter adjustment
General	Yes	Walking velocity
		Hip extension offset
Weight shift	Yes	Amplitude[a]
		Timing
		Duration[a]
		Step width
Foot clearance	Yes[a]	Knee flexion in mid swing[a]
Stance	Yes[a]	Knee flexion in mid stance[a]
Prepositioning	Yes[a]	Knee flexion in end swing[a]
Step length	Yes[a]	Step length[a]

[a]adjustable for left and right leg separately

A. User Interface

In the graphical user interface (GUI), the therapist can adjust the level of support on several aspects of gait. Therefore the gait is divided in subtasks, based on the *gait prerequisites* [44] (see Table IV). The amount of support (expressed as a percentage of the maximum support) and the reference trajectory can be adjusted for each gait subtask and each leg individually. This *selective support* [45] makes it possible to give support on only one subtask while giving complete freedom for the patient on the other subtasks, e.g., only supporting the patient in lifting his left foot during swing phase.

B. Gait Trajectory Controller

The gait trajectory controller converts the desired support levels and trajectories to joint trajectories and stiffness profiles (see Fig. 5). The gait controller consists of a Simulink model running on an embedded xPC Target PC (Mathworks Inc., Natick, MA, USA). For each joint, a trajectory is generated as a piecewise third order polynomial fitted between *key events* as described in [46]. The key event positions (timing and amplitude), and thus the trajectories, are dependent on walking velocity and patient length. The subtasks of gait are linked to

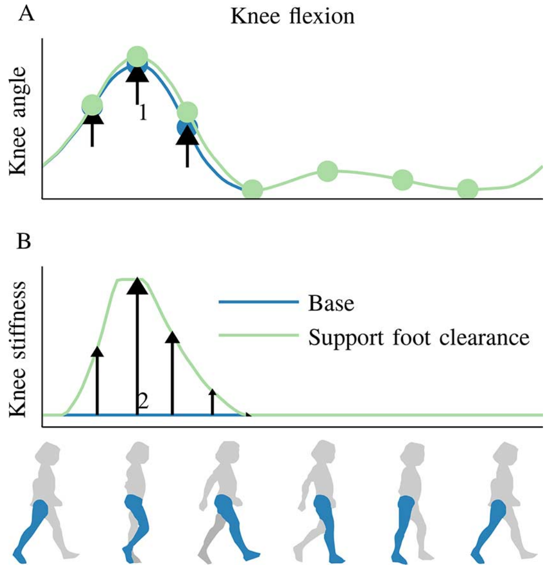

Fig. 5. Reference trajectory for the knee angle (A) and the support on the knee angle (B). Key events are plotted as dots in the angle trajectory. If the foot clearance is increased, specific key events are displaced (arrow 1), resulting in a modified reference trajectory for knee flexion in swing phase. If the support for foot clearance is increased, the stiffness of knee spring increases in swing phase (arrow 2).

the key events such that when the operator adjusts a parameter of a subtask, specific key events move relatively to their original location (see arrow 1 in Fig. 5), resulting in a modified gait trajectory for DoFs that are related to the specific subtask.

The joint trajectories are sent to the Admittance Controller which interprets them as the equilibrium position of a (critically damped) virtual spring. The spring stiffness K is related to the desired support G (in percent) by $K = K_{\max}((G)/(100))^2$ with $K_{\max}$ a predefined maximum stiffness (see Table V). The reason for the nonlinearity is that, since $K_{\max}$ is high, $(1)/(2)K_{\max}$ still *feels* very stiff; while we wanted 50% of the stiffness to feel significantly less stiff than 100% stiffness. Although algorithms exist to automatically adjust the stiffness [45], [47], we have used manually adjustable guidance as this was a specific request by the therapists.

As already stated in Section IV.A, the support can be adjusted for each gait subtask individually, resulting in a gait-phase dependent stiffness. This results in a time-varying stiffness [see Fig. 5(B)]. The real time joint angle set points and desired stiffnesses are sent to the admittance controller.

Apart from generating the gait trajectories and stiffness trajectories, the Simulink model also has various safety checks and a state machine for the transitions between different modes (self test, motors off, standing still, training, etc.).

C. Admittance Controller

The admittance controller converts the joint set points and desired stiffnesses to actuator set points. For the control of the actuators we use an admittance controller since it has the capability of displaying low and high impedance [48]. An overview of the admittance controller and its relation with the gait trajectory controller and mechanics is given in Fig. 6. In this section the components of the controller are discussed.

TABLE V
REQUIRED AND REALIZED IMPEDANCE OF LOPES II IN TERMS OF INERTIA AND DAMPING AND STIFFNESS

DoF	Required inertia	Virtual inertia	Coulomb[a]	Maximum stiffness
Pelvis AP	6 kg	40 kg	5 N	5 N/mm[b]
Pelvis ML	6 kg	40 kg	5 N	20 N/mm
Knee AP	4 kg	5 kg[c]		
Foot AP	1.5 kg	2 kg[d]		
Foot ML	1.5 kg	2 kg[d]		
Abduction			0.5 Nm	1500 Nm/rad
Thigh flexion			0.5 Nm	1500 Nm/rad
Shank flexion			0.1 Nm	
Knee flexion				1500 Nm/rad

[a]Implemented as a virtual viscous damper with a maximum force
[b]The controller is capable of rendering higher stiffness (>20 N/mm), but we chose 5 N/mm since this felt more in balance with the stiffnesses of the other DoFs
[c]Inertia in swing phase; Inertia in stance phase is 15 kg
[d]Inertia in swing phase; Inertia in stance phase is 12 kg

The central component of the admittance controller is the mass model (M^{-1}), which converts the measured and virtual forces to a desired acceleration. The mass model of LOPES II displays virtual point inertia in Cartesian coordinates i.e., the pelvis anterior/posterior translation (AP), the pelvis mediolateral translations (ML), the knee AP translation and the foot AP and ML translation. Since the controller uses segment coordinates, the inertia matrix is a non-diagonal eight by eight matrix. This implies that forces in one DoF generate accelerations in several DoFs. For each DoF the value of the virtual inertia is automatically adjusted depending on the phase of the gait. Higher virtual mass are used in stance to assure contact stability [49], and lower virtual inertias are permitted in swing phase to allow for more transparency. The virtual inertias and dampers in LOPES II admittance controller are tuned such that the system shows no oscillations in operation (see Table V). A leg in swing phase has low virtual inertia on the foot (2 kg) and knee (5 kg), which is close to the maximum allowable inertias. The virtual inertias of the foot and the knee are increased during stance phase to 12 kg and 15 kg, respectively. Since the accelerations of the foot and the knee are relatively low in stance, this increased virtual inertia is not perceived by the subject. For the pelvis the inertia is 40 kg, nearly seven times higher than desired. The Coulomb dampers are applied on the segments.

Haptic effects are implemented as virtual springs and dampers in the renderer. The springs are used to apply guidance forces; the dampers are used to damp oscillations. All effects can be limited in the force they exert on the virtual model. A limited damper therefore is called a Coulomb damper, since the behavior is similar to Coulomb friction when the damper is limited in force. The gait trajectory controller calculates the desired support in terms of the spring stiffness and joint trajectories.

The accelerations calculated by the mass model (a^U) require limiting to assure that the model position, velocity and acceleration (pva) does not exceed predefined limits e.g., speed limit of actuators and joint excursion limits. The limited model accelerations (a^L) are integrated to model velocities and positions.

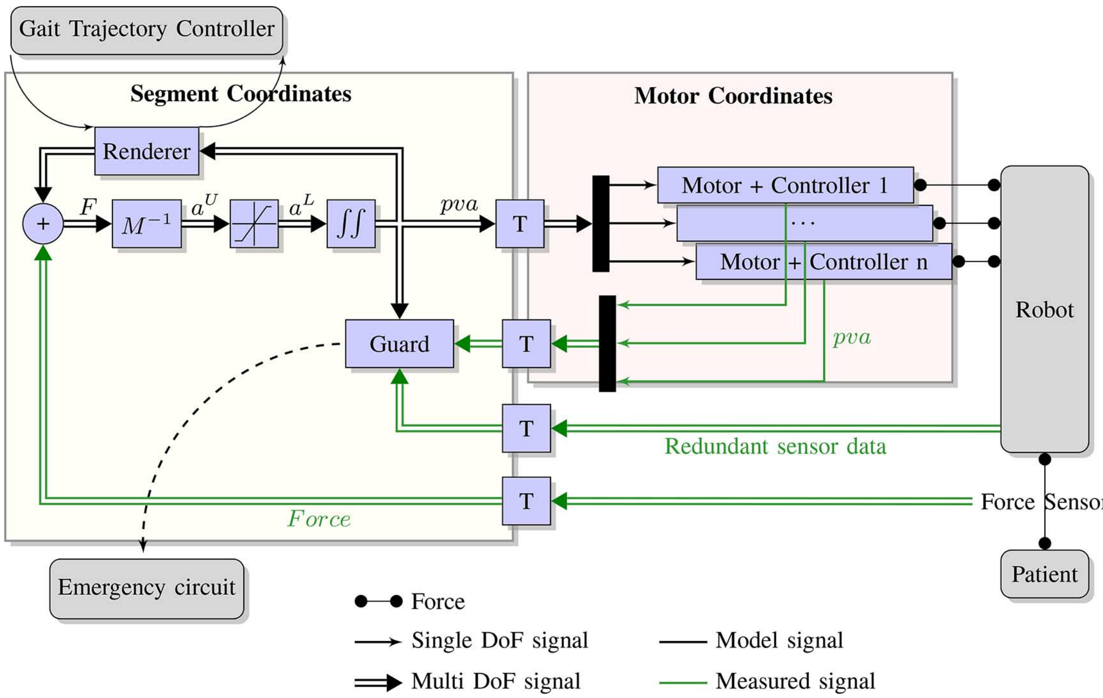

Fig. 6. Admittance controller layout of LOPES II and its relation with its peripherals. Gait controller sets the stiffnesses and positions for the guidance springs in the renderer, which calculates the supporting forces for the patient based on the spring positions and the measured positions. Sum of the renderer force and the measured force are the input for the virtual inertia (M^{-1}). Resulting model acceleration in segment coordinates (a^U) is fed through a limiter to assure the model positions, velocities and accelerations (pva) stay within bounds. Model pva are transformed (T) to motor coordinates to serve as set points for the individual motor controllers, which control the robot, which interacts with the patient. Guard triggers the emergency circuit if the errors between the measured pva and model pva, and the errors between measured motor angles with redundant sensor angle data.

The model pva serves as setpoints for the motor controllers. The actuators and sensors in LOPES II are not directly coupled to the patient segments or joints like in most exoskeletons, but through a structure of rods. This results in a non-linear relationship between sensors and motor pva on the one hand and patient segment pva on the other hand. The Admittance Controller performs real-time transformation (T blocks in Fig. 6) between motor and sensor data (force, position), and segment data. Subsequently each actuator has its own controller which controls the actuators in velocity mode, with a bandwidth of >100 Hz.

The guard compares the model pva with the measured pva. When the difference between model and measured pva exceeds predefined limits, the guard triggers the emergency circuit to stop the system. Similarly the guard compares the measured motor angles with redundant sensor angles to assure validity of the motor encoders.

V. Evaluation and Results

A. Position Accuracy

Since the position sensors of LOPES II are not collocated with the patient's segments, we want to verify whether the patient's segment angles are calculated correctly. Structural compliance may harm the accuracy of the calculated segment angles. We used an optical tracking system (Visualeyez VZ4000, PTI, Burnaby, Canada) to determine the accuracy of segment positions and angles calculated by LOPES II based on the motor angles (LOPES controller data). We applied cluster markers (frames with three markers) on the feet, lower legs, upper legs and sternum. Individual markers were put on the knee (lateral epicondyle) and hip (greater trochanter). We put additional markers on mechaniclal structure close to the patient i.e., the leg guidance [see Fig. 1 (3)] and on the rods that are connected to the pelvis.

Two healthy subjects walked at two speeds (1.5 km/h and 2.5 km/h) in LOPES II with different support levels (0%, 10%, and 100%). The optical tracking data was sampled at a rate of 90 Hz. In post processing data was filtered for spikes (50 mm), gaps up to 30 samples were interpolated, data was low-pass filtered with a second-order Butterworth filter at 10 Hz. The LOPES controller data (recorded at 1024 Hz) and optical data were re-sampled to 100 Hz, synchronized and cut into steps. Segment angles and positions were calculated from the positions of the markers on the subject (subject marker data). We also calculated the segment angles and positions from the markers on the LOPES structure (LOPES marker data).

To assess the position accuracy we calculated the root mean square error (RMSE) between LOPES controller data and LOPES marker data. Subsequently we calculated the position accuracy up to the clamps i.e., the RMSE between the LOPES controller data and LOPES marker data, to assess the inaccuracy caused by the control loop and mechanical structure (e.g., free play and mechanical compliance).

The RMSE between LOPES controller data and subject marker data is 1°–2° for the segment rotations and 7–8 mm for pelvis translations (see Fig. 7). Although the subjects are firmly strapped in LOPES II, a great part of the error can be attributed to the clamps and human tissue: for the pelvis ML translation,

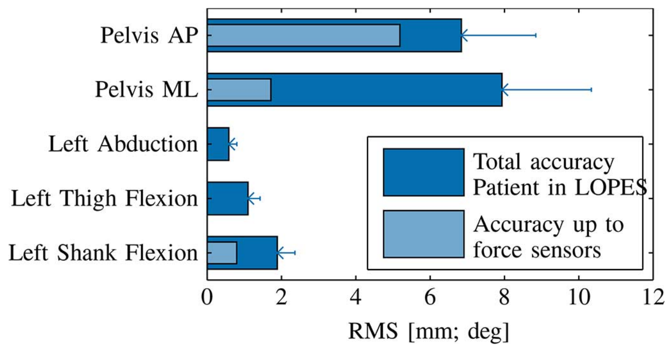

Fig. 7. Dark blue: Root mean square error between LOPES controller data (segment angles and position calculated by LOPES II) and angles and positions derived from markers on the subject; Light blue: RMSE between LOPES controller data and markers on the LOPES structure (near the clamps).

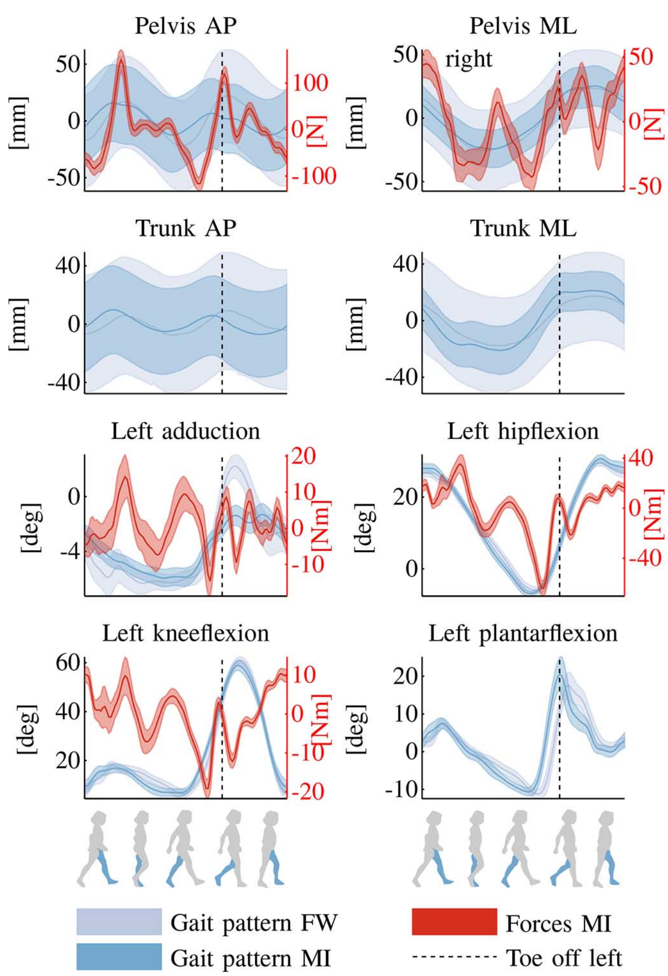

Fig. 8. Gait pattern of subject 2 at 2.5 km/h free walking outside LOPES II on a treadmill (FW) (light blue), and in LOPES II with minimal impedance (MI) (dark blue); Interaction forces in MI are plotted red; Toe-off left is indicated by the dashed line; at the left of the toe-off line, the left leg is in stance; at the right of the toe-off line, the left leg is in swing.

the position accuracy up to the clamps is 2 mm. Therefore we conclude that the compliance of the harness and human tissue are the main contributors of the position inaccuracy. In AP direction, the position accuracy up to the harness is 5 mm. In AP direction, the connection is much stiffer than in ML direction, and only accounts for 2 mm of the inaccuracy. For the shank, the accuracy up to the clamps is 1°, which is half of the total position accuracy of the shank rotation.

TABLE VI
CORRELATION (TOP) AND RMSE (BOTTOM) OF GAIT PATTERNS BETWEEN FREE WALKING AND MINIMAL IMPEDANCE WALKING FOR TWO SUBJECTS AT TWO SPEEDS

	S1		S2	
	1.5 km/h	2.5 km/h	1.5 km/h	2.5 km/h
Pelvis AP	0.84	0.33	0.91	0.62
Pelvis ML	1	0.82	0.96	0.99
Trunk AP	0.75	0.61	0.86	0.35
Trunk ML	1	0.80	0.93	0.95
Left abduction	0.95	0.84	0.93	0.88
Left hipflexion	1	0.98	0.99	1
Left kneeflexion	0.99	– [a]	0.99	0.99
Left plantarflexion	0.87	– [a]	0.9	0.82
Pelvis AP [mm]	12.3	11.9	9.2	10.9
Pelvis ML [mm]	3.4	11.0	7.2	2.7
Trunk AP [mm]	4.3	5.0	4.9	5.7
Trunk ML [mm]	11.5	12.9	14.4	5.7
Left abduction [deg]	1.3	1.6	1.6	1.5
Left hipflexion [deg]	1.0	2.6	1.2	1.4
Left kneeflexion [deg]	2.4	– [a]	2.3	2.7
Left plantarflexion [deg]	3.5	– [a]	3.7	4.6

[a]Too many missing markers on the left shank

B. Minimal Impedance

The minimal impedance mode (MI) of LOPES II (0% support) was evaluated by comparing gait patterns of the subjects walking in minimal impedance mode in LOPES II with free walking (FW) on a treadmill. We used the same marker layout and walking velocities as described in Section V-A.

Gait patterns in minimal impedance mode resemble gait patterns of free walking (Fig. 8). For the joint angles and trunk and pelvis ML motion, the correlation between minimal impedance and free walking is high (>0.8, see Table VI), and the RMSE of the difference of the gait patterns is a few degrees. For trunk and pelvis AP motion the correlation is lower, especially at higher speeds.

An explanation for the difference in correlation between pelvis AP and ML motion can be found in the acceleration. The acceleration in AP direction is higher [31] and consequently the interaction forces (needed to accelerate the virtual inertia) are higher. This is confirmed by the force patterns in Fig. 8 and the peak-to-peak values of the interaction forces (see Table VII). For the joints the interaction torques are considerably lower during swing than during stance (see Table VII). Although the accelerations of the swing leg are higher than the accelerations of the stance leg, the virtual inertia during swing is considerably lower than during stance (see Table V), and therefore the interaction forces are lower.

C. Donning Time

We recorded the donning time for several stroke patients $(N = 13)$ with Functional ambulation category (FAC) scores ranging from 0 to 4. The donning procedure for patients who

TABLE VII
PEAK-TO-PEAK INTERACTION FORCES/TORQUES IN MINIMAL IMPEDANCE WALKING FOR TWO SUBJECTS AT TWO SPEEDS. FOR ABDUCTION, HIP FLEXION AND KNEE FLEXION, THE INTERACTION TORQUES ARE SPLIT IN SWING PHASE (SW.) AND STANCE PHASE (ST.)

		S1		S2	
		1.5 km/h	2.5 km/h	1.5 km/h	2.5 km/h
Pelvis AP [N]		107.6	169.2	125.7	266.8
Pelvis ML [N]		55.2	76.1	52.0	87.6
Left abduction [Nm]	Sw.	3.3	7.8	5.4	18.1
	St.	14.2	13.6	8.4	28.9
Left hipflexion [Nm]	Sw.	20.6	33.3	21.9	39.8
	St.	32.9	69.1	41.3	99.3
Left kneeflexion [Nm]	Sw.	8.4	8.7	12.8	22.6
	St.	9.0	21.1	14.2	29.2

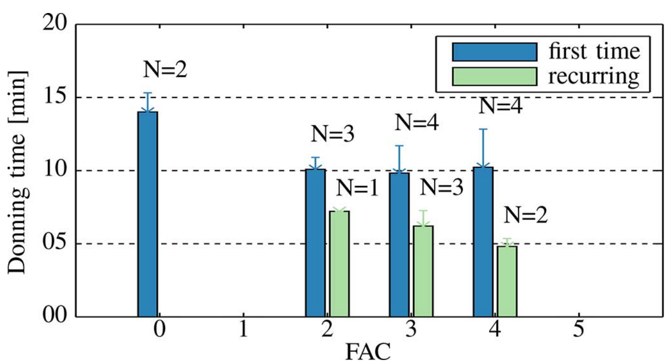

Fig. 9. Donning time grouped by FAC score and separated in first training and recurring training. For FAC 0 a lift was used to transfer the patients from wheelchair to LOPES II.

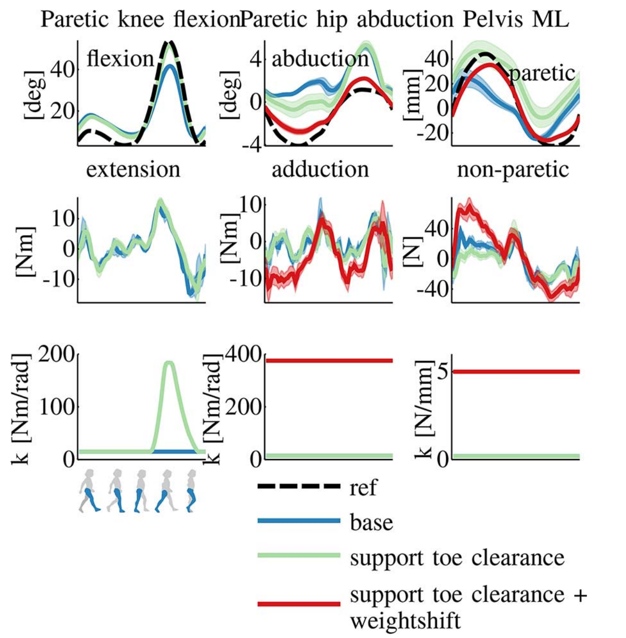

Fig. 10. Assist-as-needed with a chronic stroke survivor (FAC 5), 1.5 km/h. Joint angles, interaction torques and virtual spring stiffness are shown for different support conditions. The spring stiffness is denoted by "k," the spring's reference trajectory denoted by "reference." For clarity we omitted the support toe clearance plus weight shift (red line) for knee flexion plots, since the lines are similar to the lines of support toe clearance.

perform training with LOPES II for the first time consists of five steps: 1) the therapist measures the length of the upper leg, lower leg and foot length; 2) the measured data and other patient data e.g., weight, posture is fed into computer; 3) the patient is prepared to get into LOPES II i.e., getting to stand from wheelchair, optionally apply a sling for the paretic arm and put into standing position. For patients with FAC 0 the harness is applied in the wheelchair and the body weight support (BWS) system is used to lift the patient out of the wheelchair into LOPES II. Additionally the leg guidance (Fig. 1) is set to the length of the lower leg; 4) the therapist firmly straps the patient in LOPES II. This is done while the patient is standing, if needed the BWS is used; 5) LOPES II is put in an active mode from which the training can start.

For recurring training, steps one and two of the donning procedure are not needed, since the settings are stored in the database, resulting in a shorter donning time. Therefore we did a recording of the recurring donning time for six patients.

The average first donning time is 11 min and 2 s. The average recurring donning time is 6 min and 4 s (see Fig. 9). For patients with FAC 2–4 the donning time for first training meets the goal of 10 min. For patients with FAC 0 donning time was longer (14–15 min), due to the use of the lift. For recurring trainings (no limb measurement needed) the donning time was 5–8 min. This approaches the desired donning time of 5 min for recurring patients. A limitation is that no data was available of recurring training of FAC 0 patients. There seems to be a trend that a higher FAC score shortens the donning time, but there were too few measurements to statistically confirm this trend.

D. Pilots With Patients

We performed exploratory studies with stroke survivors and SCI patients. We will discuss two extreme cases: a mildly impaired stroke survivor (FAC 5) and a severely impaired SCI patient (lesion level C1, FAC 0).

A stroke survivor (FAC 5) walked in LOPES II at 1.5 km/h, first with 10% support and 0% BWS. The subject showed a stiff-knee gait on the right leg and used a little circumduction (5° abduction) as compensation strategy. Subsequently we applied selective support on the foot clearance, i.e., support (high stiffness) on the paretic knee flexion during the swing phase. The paretic knee showed an increased knee flexion from 42° to 51° (see Fig. 10). Despite the increased support in knee flexion during the swing phase, the interaction torque did not increase. This can be attributed to an intuitive response of the subject to minimize the interaction force. Though the support occurred only during the swing phase, the pelvis ML translation increased to the paretic side and the paretic adduction increased during stance phase of the paretic leg. This indicates that the subject took more weight on the paretic leg, during support on toe clearance.

Next we added selected support on the weight shift. This means that the stiffness for pelvis ML and for the abduction/adduction of both legs was increased for the complete gait cycle. The pelvis ML motions increased (see Fig. 10). Contrary to the support on foot clearance, the subject did not minimize the interaction forces. Due to the increase in stiffness in abduction, the

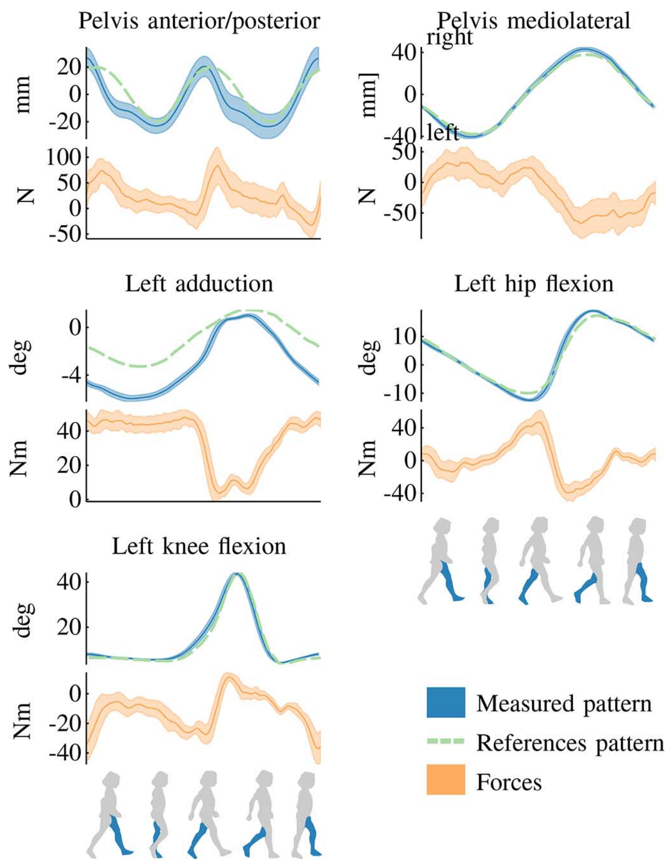

Fig. 11. Gait pattern and interaction forces of a 114 kg SCI patient (FAC 0) walking in LOPES II with 80% support and 40% bodyweight support at 0.7 km/h.

paretic abduction decreased to 2°, partially canceling the circumduction.

A SCI patient (FAC 0; 114 kg) walked in LOPES II with a general support of 80% and 40% bodyweight support (BWS) at 0.7 km/h (see Fig. 11). The patient's joint angles and pelvis translations followed the reference trajectories. Since the support is implemented as springs, the interaction forces are largely proportional to the tracking errors. This is reflected in the pelvis AP force: the subject was leaning backward (negative interaction force). Consequently the measured trajectory of pelvis AP was a little behind its reference trajectory. This also resulted in more abduction of the left leg, especially during left stance phase: interaction torque of -40 Nm and a tracking error of 2°. In swing the subject is able to follow the reference trajectory: the interaction torque and tracking error approach zero.

VI. Discussion

In this paper we discussed the design, the technical validation and the preliminary evaluation with patients of a shadow leg based robotic gait trainer with admittance control. We demonstrated that LOPES II is suitable for Assist-As-Needed training and that the required donning time is acceptable for clinical practice.

LOPES II uses the new shadow-leg concept. In this concept the motors are not collocated with the patient's segments, but are coupled to the segments with a structure of rods and levers. The mechanics are located behind the patient. This implies that mildly impaired patients have the freedom to walk with arm swing; the more severely impaired, can use the side railing to aid in balance support.

Despite this complex structure, LOPES II calculates the subject segment angles with an accuracy of $\pm 2°$. The position accuracy up to the clamps is better and therefore a considerable part of the error can be attributed to the compliance between the robotic structure and the subject (see Fig. 7). A stiffer connection between the LOPES structure and the subject's skeleton would improve the agreement between LOPES data and the subject marker data, and consequently the position accuracy. The most compliant elements between LOPES II and the subject's skeleton are human tissue and the harness. In other words, if the position accuracy should be improved the main focus should be on stiff clamping. At the pelvis translations the accuracy is 7–8 mm. Here the effect of the compliance of the harness is considerable especially in the ML direction. A stiffer clamping of the harness, will improve the position accuracy considerably.

Neckel *et al.* [50] investigated the position accuracy of the Lokomat in terms of Cartesian joint positions, by comparing optical tracking data with Lokomat data. They found an average error (offset), projected on the sagittal plane, of 12 mm for the left hip and 18 mm for the right hip. Assuming a 400 mm upper leg, this results into a hip flexion error of 1.7°–2.6° for the Lokomat. Therefore we conclude that the position accuracy of LOPES II, despite the non-collocated motors with the complex structure, is similar to that of Lokomat.

To cover the complete spectrum from low to high impedance, we used an Admittance Controller. The minimal impedance of LOPES II is reflected by an inertia (see Table V). This means that the patient has to apply force to LOPES II to accelerate the virtual inertia. If the virtual inertia is sufficiently low, the forces the patient has to apply are negligible. In an earlier study we found that gait patterns are not noticeably affected when an inertia of less than 2 kg is added to the ankle [31]. For the ankle motions, the achieved inertias in swing-phase are the required 2 kg, whereas in stance phase, the inertias are considerably higher than required to maintain contact stability. This has minimal effect on free walking, since the leg has a high impedance during stance. Similarly for the knee, the admittance controller adds 5 kg of inertia during swing phase, whereas we estimated that 4 kg was allowed.

The scaling of the inertia is reflected in the interaction forces. During stance the interaction forces are considerably higher than in swing (see Table VII). We believe that for transparency, the interaction forces during swing phase are more important, since the impedance of the leg is lower, and the desired accelerations of the leg are higher. The range of interaction torques of LOPES II in swing are 8–23 Nm for the knee flexion, and for the hip flexion 20–40 Nm. The torques increase with speed, but also show large differences between the two subjects.

Reference material for interaction torques is scarce. For the Lokomat in zero impedance mode, Riener *et al.* [11] reported interaction torques of 38–42 Nm for the hip and 25–28 Nm at the knee at a speed of 2 km/h. Van Dijk *et al.* [51] experimented on reduction of the interaction forces on Lopes I. The baseline peak-to-peak interaction torques at the hip and knee are 6 Nm and 17 Nm, respectively. Using dynamics compensation and acceleration feed-forward, the RMS of the interaction forces at the thigh reduced with 35%–39%, but at the shank the change in interaction force was negligible. Direct comparison of our results with these two studies is difficult. In the Lokomat study the interaction torques are not recorded directly, but reconstructed from forces measured at the suspension of the actuators. Van Dijk *et al.* [51] did record direct interaction forces, but they did not convert the forces to joint torques. The joint torques that are reported are measured in the series elastic actuators. Dynamics of the mechanics between the SEA and the subjects negatively affect the interaction torques. Therefore we conclude that LOPES II has lower interaction torques than the Lokomat and higher interaction torques than Lopes I.

A second criterion for transparency is how much gait patterns in minimal impedance mode resemble the patterns of free walking. The joint angle patterns in minimal impedance mode are similar to free walking (see Fig. 8). This is reflected in a high correlation (0.84–1) and a low RMSE of a few degrees on the joint rotations which is similar to the normal intra-subject-variability of normal walking [46]. However for pelvis and trunk AP translation the deviations in amplitude and phase are larger, resulting in lower correlations (0.33–0.91). This is not surprising considering the fact that the inertia (40 kg) is nearly seven times higher than required (6 kg [31]). The RMSE values are considerable (4–12 mm), however they are less than the within-subject standard deviation of the patterns (see Fig. 8).

For the pelvis- and trunk ML the correlations are high (0.8–1), despite the fact that also for the pelvis ML translation, the inertia is seven times higher than required. This is explained by the fact that the acceleration in ML direction are lower compared to AP direction [31], and consequently the inertial forces are lower (see Fig. 8). Therefore the transparency in ML direction is higher than in AP direction. However, still the RMSE values in ML direction are considerable (3–14 mm), indicating that there is room for improvement of the transparency.

Based on the relatively low interaction forces and the high correlation of gait patterns with free walking, we conclude that LOPES II has a reasonable transparency in minimal impedance mode. In the minimum impedance mode, LOPES II displays an inertia (non zero impedance) and therefore the interaction forces are required to accelerate this virtual inertia. The transparency may be improved further by reduction of the virtual inertia, especially in the pelvis AP direction, or to provide a feed forward on the acceleration of the virtual inertia [51].

For clinical application of robotic aided gait training it is paramount that the donning time is short. The donning time for LOPES II is 10–15 min for first training (see Fig. 9). Recurring donning time is 5–8 min for patients FAC 2–4. For the severely impaired, the first donning time and probably also the recurring donning time (although not measured) are higher than desired. This can be attributed to the fact the donning severely impaired patients includes the process of lifting out of the wheelchair. The donning time for the more severely affected patients can be shortened by improving the process of lifting the subject out of the wheelchair.

The goal was to provide robotic gait training for a wide range of patients from mildly to severely impaired. LOPES II was shown to be powerful and stiff enough to enforce a walking pattern on a severely affected patient (SCI lesion level C1; FAC0; 114 kg). We also demonstrated that, on the other side of the spectrum, LOPES II can provide selective support to a mildly affected patient (FAC5). This resulted in the anticipated effect on the supported aspect of gait. The patient seemed to adapt to support on toe clearance since the interaction torques did not change. Remarkably the stroke survivor also showed minor changes in aspects of walking that were not supported directly. We assume that when subjects receive selective support, they adapt different aspects of their gait pattern, also aspects that are not supported directly, to find a new optimal gait pattern. For this process the minimal impedance of LOPES II is paramount, since it gives the patient the freedom to adapt his gait pattern. This showed that LOPES II is capable of providing Assist-As-Needed.

VII. Conclusion

The robotic gait trainer LOPES II has been designed to meet the requirements stated by physiotherapists, patients, researchers and rehabilitation physicians. The main goals were to facilitate Assist-As-Needed training and realize short donning time. We have built a device with has eight admittance controlled DoFs to cover the spectrum of low and high impedance. Additionally free motion is allowed in the remaining DoFs of gait. The mechanical structure of the shadow-leg does not require exact joint alignment and uses a minimal amount of clamps. This allows for a short donning time. LOPES II can apply maximum support to severely impaired patients and minimum support to healthy subjects. With the gait controller selective support can be applied to specific aspects of gait. We conclude that LOPES II is suitable for Assist-As-Needed training. This makes LOPES II promising for clinical application. Clinical studies are needed to answer the questions how gait training with the LOPES II compares with conventional therapy or other gait training robots, in terms of effectiveness and cost.

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