

A Novel Mechatronic Body Weight Support System

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Abstract—A novel mechatronic body weight support (BWS) system has been developed to provide precise body weight unloading for patients with neurological or other impairments during treadmill training. The system is composed of a passive elastic spring element to take over the main unloading force and an active closed-loop controlled electric drive to generate the exact desired force. Both force generating units, the passive spring and the active electric drive, act on the patient via a polyester rope connected to a harness worn by the patient. The length of the rope can be adjusted with an electric winch to adapt the system to different patient sizes. The system is fully computer controlled. At unloading loads of up to 60 kg and walking speeds of up to 3.2 km/h, the mean unloading error and the maximum unloading error of the presented BWS system was less than 1 and 3 kg, respectively. The performance was compared with those of two purely passive BWS systems currently being used by most other rehabilitation groups. This comprised counterweight systems and static BWS systems with fixed rope lengths. Counterweight systems reached mean and maximum unloading errors of up to 5.34 and 16.22 kg, respectively. The values for the static BWS were 11.02 kg and 27.67 kg, respectively. The novel mechatronic BWS system presented in this study adjusts desired unloading changes of up to 20 kg within less than 100 ms. Thus, not only constant BWS, but also gait cycle dependent or time variant oscillations of the desired force can be realized with high accuracy. Precise and constant unloading force is believed to be an important prerequisite for BWS gait therapy, where it is important to generate physiologically correct segmental dynamics and ground reaction forces. Thus, the novel BWS system presented in this paper is an important contribution to maximize the therapeutic outcome of human gait rehabilitation.

Index Terms—Dynamic body weight support (BWS), force control, ground reaction force, rehabilitation of neurological impairments, rehabilitation robotics, treadmill training.

I. INTRODUCTION

A. Rationale of Body Weight Support

BODY weight support (BWS) systems are increasingly being used in conjunction with treadmills to assist therapists in gait training of patients with neurological impairments.

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Paralysis of the leg muscles often prevents patients from supporting their own body weight during stance phase. Reducing the gravitational forces acting on the legs by a BWS system would reduce the load that needs to be overcome by the patient and facilitate stepping movements. In addition, the BWS system ensures safety and stability of the patient walking on the treadmill.

The body weight unloading is high in the beginning of training (60%–80% support of the body weight). As the mobility of the patient improves with time, the amount of BWS is reduced. By gradually decreasing the amount of BWS provided to the patient, afferent signals to gait-related brain centers increase. In this way, gait “retraining” is supported by the appropriate activation of sensory receptors at the appropriate times during the gait cycle.

There are numerous secondary benefits to gait retraining. A prolonged period bound to a wheelchair or bed can lead to many medical and psychosocial problems. Some of the most common problems are pressure sores, reduced bone-density in the legs, increased risk of fractures, deterioration of cardiopulmonary and circulatory functions, spasticity, bowel and bladder stagnation, urinary tract infections, and joint contractures. The exercise and mobility provided with treadmill training offers some prophylaxis to these possible problems.

B. Overview of Body Weight Support Systems

Body weight support systems applied to gait therapy normally consist of a harness system worn by the patient, ropes and pulleys, and a counterpoise to unload the patient. There exist many different counterpoise systems, as follows.

1) *Static Systems*: In static systems, the harness is attached to an overhead suspension, usually a system composed of ropes and pulleys connected to a winch [Fig. 1(a)]. The winch actuation can be realized manually [1], or automatically by an electric drive [2] or a fluidic system [3]. The winch is wound up until the rope supports the desired unloading. More advanced systems use a force transducer that measures the weight support during hoisting. The drive stops as soon as the desired patient unloading is reached. The main disadvantage of static systems is the limitation of vertical movement of the center of mass. Thus, the constancy of the patient unloading during walking is not guaranteed and the execution of a physiologic gait pattern may be hampered.

2) *Passive Counterweight Systems*: A counterweight can be used to dynamically unload part of the patient’s body weight. The counterweight is connected to the patient harness by a rope-and-pulley system [Fig. 1(b)] [8]. However, the counterforce comprises not only the counterweight’s gravitational force but also an inertial component, resulting from accelerations of the counter mass when moving the body up and down during walking. These inertial forces cause deviations of the

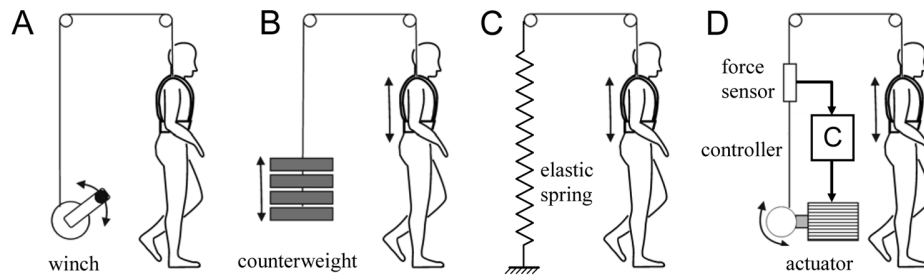


Fig. 1. Illustration of the most common body weight support (BWS) systems. (A) Static BWS with rope and winch. (B) Passive dynamic BWS with adjustable counterweights. (C) Passive dynamic BWS with elastic spring. (D) Active dynamic BWS with force sensor and electric winch.

suspension force from the desired value. Further disadvantages of common counterweight systems are that only discrete force values can be chosen and that the adjustments have to be done manually.

3) *Passive Elastic Systems*: Elastic components such as a metal spiral spring or a bungee cord can be used to unload the subject [Fig. 1(c)] [4]–[6]. The amount of unloading is determined by the amount of tension of the elastic element. Compared to counterweights, the advantage of this approach is that inertial effects can be neglected. However, the suspension force is a function of spring length and, therefore, varies due to the vertical movement of the patient. Force variation can be reduced using a spring with low stiffness. This requires a long spring elongation in order to obtain the required working load. Kram *et al.* [6] obtained rather large elongations using a series of springs connected by cables and separated by pulleys. However, in such systems, it is difficult to allow load adjustments over a large range.

Passive elastic unloading can also be achieved with a closed pneumatic cylinder filled with compressed gas [7] instead of elastic springs. Gazzani *et al.* [8] proposed a system, where a closed-loop controller is used in order to compensate for small pressure losses due to air leakage.

4) *Active Dynamic Systems*: BWS can also be realized with active dynamic systems in which a pneumatic, hydraulic, electromagnetic, or any other active force-generating unit produces the desired force during walking [Fig. 1(d)]. The vertical position or supporting force to the patient's movement is adjusted by applying a closed-loop approach. Vertical position or force is measured by a position sensor or force sensor, respectively, and fed into a controller that sends commands to the actuator to generate the desired positions or forces.

Several groups developed closed-loop systems that are based on pneumatic actuators [1], [9], [10]. Pneumatic systems can be made to behave rather compliant due to the compressibility of air, thus providing force constancy during small movements. A closed-loop control ensures force constancy over time and during large vertical movements. However, bandwidths are limited to less than 20 Hz, which is insufficient to high performance tasks [11].

In contrast, hydraulic actuators have very high output stiffness, which make the actuator essentially a pure position source. Thus, force control needs to be realized with a position based force controller, which needs very fast actuators to ensure a high quality force control [12]. In general, this type of closed-loop

controller generating a constant force while compensating friction and inertial forces tends to instability [13]. Whereas Gordon *et al.* [14] have suggested the use of hydraulic actuators, their design has not yet been realized.

Electromagnetic actuators can either be used as a position source in combination with a position sensor and a closed-loop position control or as a force source in combination with a back-drivable transmission. Thus, they comprise the advantages of both fluidic and pneumatic actuators. However, only rough concepts have been presented in the literature with no or only preliminary experimental results [15], [16].

C. Objective of This Work

The main mechanism to conserve the energy expenditure during locomotion is the exchange between kinetic and gravitational potential energy within each stride [17]. The percentage of energy recovery reaches about 65% for the optimal walking speed. In this way, the muscle work against gravity to accelerate the center of mass is minimized. This mechanism is wholly preserved if the applied unloading force is constant over the stride period. In this case, the physical work spent against the unloading force is zero. This is similar to walking at reduced gravity.

This is not the case if the unloading force is not constant as the physical work against the unloading force is the integral of the unloading force multiplied by the vertical speed over time. The energy saving mechanism mentioned above may be significantly disturbed. This may affect the time course of the movement, joint moment, and ground reaction forces. However, constancy of unloading force is an important prerequisite for BWS gait therapy, where it is important to generate physiologically correct segmental dynamics and ground reaction forces.

A novel mechatronic body weight support system for precise unloading during treadmill walking is presented in this publication. The design goals for the novel BWS system were as follows:

- high force accuracy with a force error clearly lower than the error of alternative BWS systems available on the market;
- full automation with static lifting capacity of up to 150 kg;
- dynamic unloading capacity of up to 80 kg.

The mechanical setup and the closed-loop control system are presented in detail. This is followed by a validation of the device among two different BWS systems currently on the market.

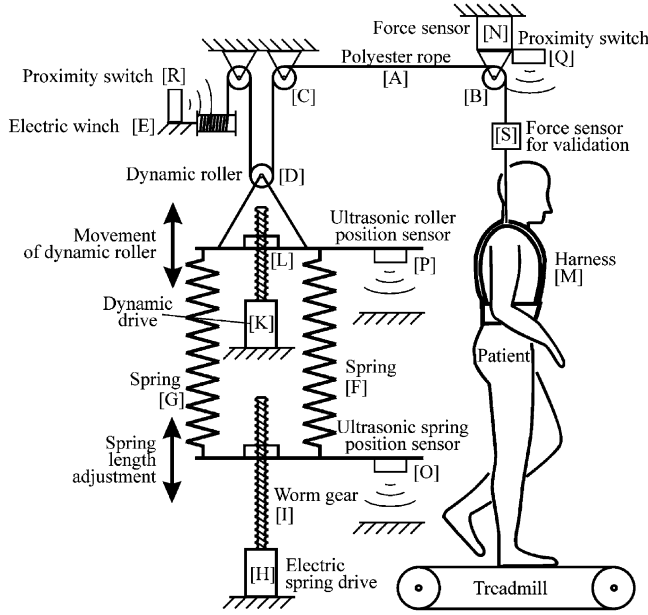


Fig. 2. Sketch of the mechanical setup of the Lokolift system.

II. METHODS

A. Mechanical and Electrical Setup

The BWS system presented is called “Lokolift” and combines the key ideas of both passive elastic and active dynamic systems. The patient is placed in a harness, which is connected to the BWS system via an 8-mm-diameter polyester rope (Fig. 2[A]). The material and the diameter of the rope is a result of extensive testing procedures performed during the design phase of the Lokolift system (see Section II-D). The rope is guided via two static pulleys seated by ball bearings (Fig. 2[B, C]). The rope enters the BWS system in the vertical direction and is guided around a pulley (Fig. 2[D]) that is capable of moving up to 7 cm in the vertical direction. The end of the rope is attached to an electric winch (Fig. 2[E]), which serves to lift the patient from a sitting position as well as to keep the system in an optimal working range. The winch is capable of lifting loads of up to 150 kg. The vertical displacement of the moving roller (Fig. 2[D]) can cause patient lifting movements in the range of 14 cm, because the pulley (Fig. 2[D]) is part of a tackle. Accordingly, half of the load applied to the moving roller (Fig. 2[D]) acts on the patient. Two elastic springs (Fig. 2[F, G]) with a spring stiffness of $k_{\text{spring}} = 2 \text{ kN/m}$ are connected to the moving roller. The lower ends of the springs are attached to a worm gear (Fig. 2[I]). The spring length and, thus, the prestress of the springs can be adjusted by an electric drive (Fig. 2[H]) (maxon RE 40, maxon motor AG, Switzerland). The force f_{spring} of a single spring also acts on the patient body as long as the moving roller (Fig. 2[D]) is not blocked by one of its stop limits. A third electric drive unit (Fig. 2[K]) is mounted in parallel to the springs. It includes another maxon motor RE 40 with a custom-made ball screw (August Steinmeyer GmbH & Company KG, Germany) (Fig. 2[L]) as it is used within the actuated gait orthosis Lokomat [18]. The drive unit can generate a peak force of up to 1500 N. The force f_{drive} generated by the dynamic drive (Fig. 2[K]) is applied to

the moving roller and, thus, half of this force acts on the patient fixation (Fig. 2[M]) as long as the drive is not at its stop limits. Neglecting friction, gravity, and dynamical effects, the patient force f_{patient} can be described by

$$f_{\text{patient}} = \frac{2 \cdot f_{\text{spring}} + f_{\text{drive}}}{2}. \quad (1)$$

The system is equipped with a strain gauge based force sensor (Fig. 2[N]) (Baumer Electric AG, Switzerland) to measure the force at the static pulley (Fig. 2[B]) above the patient. The force sensor is compensated for shear loads in medial-lateral direction. The unloading force f_{patient} is calculated from the sensor force f_{sensor} as

$$f_{\text{patient}} = \frac{f_{\text{sensor}}}{2 \cos\left(\frac{\alpha}{2}\right)}$$

(see also Fig. 2).

The patient force will be erroneous if the subject is not walking below the force sensor. In this case, an additional force sensor (Fig. 2[S]) integrated into the rope can be used to measure the actual unloading force.

One ultrasonic position sensor (Baumer Electric AG, Switzerland) (Fig. 2[O]) measures the length of the springs. A second sensor detects the position of the moving roller (Fig. 2[P]). One proximity (Fig. 2[Q]) switch supervises the distance between the patient fixation and the sensorized pulley (Fig. 2[B]) to avoid collision of the patient fixation with the pulley (Fig. 2[B]). Another sensor (Fig. 2[R]) is attached to the winch (Fig. 2[E]) and generates a warning signal as soon as the rope is wrapped less than twice around the winch to avoid that the rope fully decoils from the winch.

All analog sensor signals are filtered with a first order antialiasing low-pass filter before being digitized by an analog-to-digital converter (CIO-DAS16, Measurement Computing, Norton, MA). A digital-to-analog card (CIO DDA06, Measurement Computing, Norton, MA) operates three alternating current servo controllers (ADS series, maxon motor AG) of the electric motors. The whole system is controlled by an IntelP4 processor at 2.8-GHz clock speed operated with the xPC Target (MathWorks Inc., Natick, MA) real time kernel (Fig. 3).

The control software for the entire BWS system is programmed with Matlab Simulink (The MathWorks Inc., Natick, MA). A real-time code for operation at the xPC Target computer is automatically generated by the Real-Time Workshop. The user interface is programmed with LabView (National Instruments, Austin, TX) and runs at a second host PC connected to the target PC via Ethernet.

B. Model of the Dynamic Drive of Lokolift

The dynamic drive of Lokolift was modelled to derive a suitable control architecture explained in the proceeding paragraph. The drive for adjusting the spring length (Fig. 2[H]) and the winch (Fig. 2[E]) are not considered in this dynamic model, because their effects to the unloading are quasi static compared to the movements of the dynamic drive. The Lokolift dynamics

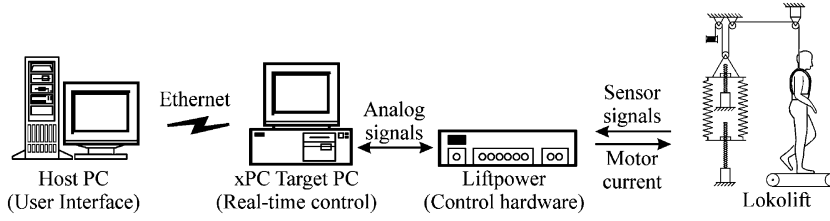


Fig. 3. Control hardware of the Lokolift system. User interface programmed with LabView runs at the host PC. Real-time xPC Target is equipped with analog-digital and digital-analog converter cards. Liftpower module is equipped with servo controllers and electric circuits for signal preprocessing.

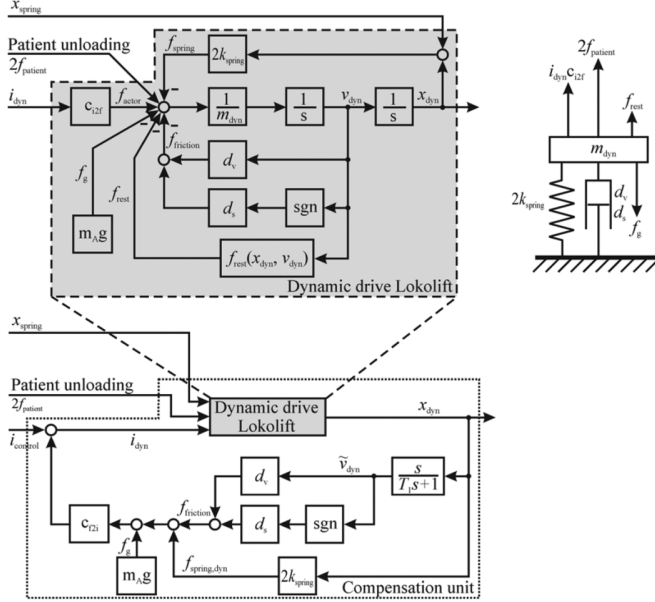


Fig. 4. Block diagram of the dynamic Lokolift drive model (top) and the active compensation unit (bottom).

was modeled as a moving mass [representing the dynamic drive (Fig. 2[K]) with pulleys (Fig. 2[B], [C], and [D])] that is coupled to the environment via a spring-damper element (representing the springs (Fig. 2[F]), as well as static and dynamic friction). Patient force f_{patient} , dynamic drive force f_{actor} , gravitational force f_g of the moving roller, and a term f_{rest} act on the moving roller (Fig. 4). The force f_{rest} comprises all modeling errors such as a part of the static friction, as well as nonlinear effects of the polyester rope (Fig. 2[A]). The dynamic equilibrium of force can be expressed by

$$2f_{\text{patient}} = i_A c_{i2f} + 2k_{\text{spring}}(x_{\text{spring}} + x_{\text{dyn}}) + d_v v_{\text{dyn}} + d_s \text{sgn}(v_{\text{dyn}}) + m_{\text{dyn}} a_{\text{dyn}} + f_g + f_{\text{rest}} \quad (2)$$

where i_A is the current of the dynamic drive (Fig. 2[K]), c_{i2f} is the constant factor between the motor current and the static force of drive (Fig. 2[K]), k_{spring} is the spring stiffness, x_{dyn} is the position of the moving roller (Fig. 2[D]) with its working range from -35 to 35 mm, x_{spring} is the spring length in neutral position of the moving roller, d_v is the viscous friction coefficient, d_s is the static friction coefficient, v_{dyn} is the velocity, and a_{dyn} the acceleration of the moving roller (Fig. 2[D]). Note that

the spring force is equal to the spring constant k_{spring} multiplied by the sum of the spring length in neutral position of the moving roller x_{spring} and the position of the moving roller x_{dyn} . The effective mass m_{dyn} comprises all moving parts of the Lokolift system such as the pulleys, the linear drive, the gear, and the armature of the electric drive. The patient force is multiplied by two, because the moving roller are part of the tackle.

C. Lokolift Controller

Assuming an ideal model ($f_{\text{rem}} = 0$) and static conditions ($v_{\text{dyn}} = 0$, $a_{\text{dyn}} = 0$), (2) can be simplified to

$$2f_{\text{patient}}(t) = i_A(t) c_{i2f} + 2k x_{\text{spring}} \quad (3)$$

for the neutral position ($x_{\text{dyn}} = 0$). One of the aims for the controller design (Fig. 5) was to minimize the electrical power of the dynamic drive and, thus, avoid thermal overload. The electric energy of the drive is

$$E = \int_t R_A (i_A(t))^2 dt = \frac{R_A}{c_{i2f}^2} \int_t (2f_{\text{patient}}(t) - 2k \cdot x_{\text{spring}}(t))^2 dt \quad (4)$$

where R_A is the electric resistor of the armature circuit of the electric drive. The energy E can be minimized for

$$x_{\text{spring}} = \frac{\bar{f}_{\text{patient}}}{k} = \frac{\frac{1}{T} \int_0^T f_{\text{patient}} dt}{k} \quad (5)$$

leading to the minimal energy value

$$E_{\text{min}} = \frac{R_A}{c_{i2f}^2} \text{var}(2f_{\text{patient}}) \quad (6)$$

where $\text{var}(f_{\text{patient}})$ is the variance of the patient unloading and T is a sufficiently large integration time. The desired spring drive position $x_{\text{spring},d}$ can be derived from (5)

$$x_{\text{spring},d} = \frac{\bar{f}_d}{k} \quad (7)$$

assuming that the patient unloading f_{patient} is equal to the desired unloading f_d . The mean desired unloading $\bar{f}_d$ is equal to f_d if the desired unloading is constant. The spring position is

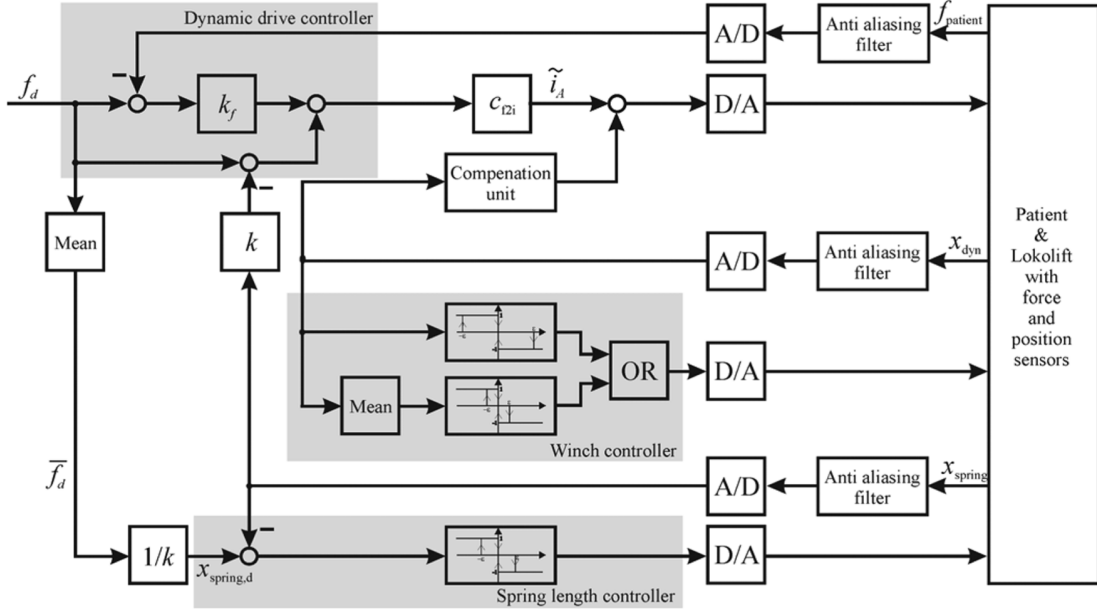


Fig. 5. Controller of the Lokolift system including all three drives.

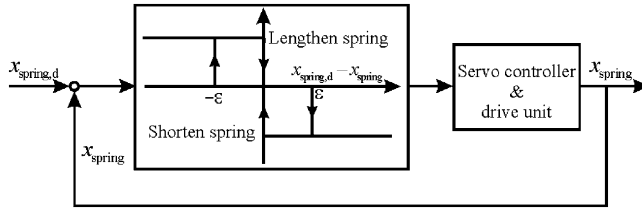


Fig. 6. Closed-loop position controller with hysteresis applied to adjust the spring length.

adjusted by a simple closed-loop position controller (Fig. 6). The spring starts shortening if $x_{\text{spring}} > x_{\text{spring},d} + \varepsilon$ and starts lengthening in the case of $x_{\text{spring}} < x_{\text{spring},d} - \varepsilon$. The spring drive stops as soon as the spring is adjusted to its desired length. The position constant ε defines the width of the hysteresis of the controller, so that the drive is insensitive to measurement noise.

During dynamic operation, gravity, friction, inertial forces, and the altering part of the spring force $k_{\text{spring}}x_{\text{dyn}}$ should not be neglected as in our static considerations above. Therefore, gravity forces, friction forces, and the position dependent spring forces were partially experimentally identified. In order to approximate the ideal model (3), they were compensated by the feedback loop (Fig. 4)

$$i_A = \tilde{i}_A - \frac{kx_{\text{dyn}} + dv_{\text{dyn}} + d_E \text{sgn}(v_{\text{dyn}}) + f_g}{c_{i2f}}. \quad (8)$$

After applying (7) and (8), the resulting model of the real system is

$$\begin{aligned} f_{\text{patient}} &= \tilde{i}_A c_{i2f} + kx_{\text{spring}} + m_{\text{dyn}}a_{\text{dyn}} + f_{\text{rem}} \\ &\approx \tilde{i}_A c_{i2f} + \bar{f}_d + m_{\text{dyn}}a_{\text{dyn}} + f_{\text{rem}}. \end{aligned} \quad (9)$$

A constant desired force f_d is equal to the patient force f_{patient} if the negative force applied by the electric drive

$-\tilde{i}_A c_{i2f}$ is equal to the sum of the inertial force $m_{\text{dyn}}a_{\text{dyn}}$ and the remaining force f_{rem} .

The remaining force f_{rem} is unknown because it contains the modeling errors and, thus, cannot be compensated. The inertial force can neither be compensated, because a proper acceleration signal is not available. Calculation of the acceleration signal from the position signal is no option due to the problem of added time delay and noise.

The remaining force f_{rem} as well as the inertial force $m_{\text{dyn}}a_{\text{dyn}}$ are measured with the force sensor positioned above the patient. The error force Δf is the difference between the desired force f_d and the measured patient force f_{patient} . A proportional closed-loop force controller with the controller constant k_f is used to reduce the error force. Additionally, there is a feedforward of the desired motor force of the dynamic drive, which is zero if the desired force is constant ($f_d \equiv kx_{\text{spring}}$)

$$\tilde{i}_A = -\frac{k_f}{c_{i2f}}(f_d - f_{\text{Sens}}) + (f_d - kx_{\text{spring}}). \quad (10)$$

Assuming an ideal current controller and a rope with infinite stiffness, the patient force is

$$f_{\text{patient}} = f_d + \frac{f_{\text{rest}} + m_{\text{dyn}} \cdot a_{\text{dyn}}}{1 + k_f}. \quad (11)$$

The following mechanical design criterion for the Lokolift system can be derived from (11). The uncompensated forces f_{rem} and the inertial force $m_{\text{dyn}}a_{\text{dyn}}$ including all moving inertia need to be as little as possible, in order to minimize the error force spoiling the constant unloading of the patient.

The electric winch of Lokolift can be activated by pressing a button in the installation mode, e.g., to lift a patient out of the wheelchair. The winch is further used to keep the dynamic drive in the optimal working range in order to minimize the electric

energy (4) also during dynamic operation. The length of the rope is adjusted in such a way that the dynamic drive energy function (3) is minimized. This is the case if (5) is satisfied. But during dynamic operation, the length of the spring $x_{\text{dyn}} + x_{\text{spring}}$ does not remain constant as assumed in (5). It changes with the position of the moving roller (1)

$$x_{\text{spring,dyn}} = x_{\text{spring}} + x_{\text{dyn}} \quad (12)$$

The position of the moving roller x_{dyn} and, thus, the length of the spring depends on the length of the unwrapped rope and the vertical position of the patient. The dynamic drive will move up and, thus, lengthen the spring if the patient body moves down. This effect is compensated if the rope is lengthened by the winch. As a result, the winch is operated in order to keep the average position of the moving roller $\bar{x}_{\text{spring,dyn}}$ zero

$$\bar{x}_{\text{dyn}} = \frac{1}{T_{\text{av}}} \int_{t-T_{\text{av}}}^t x_{\text{dyn}} dt = 0. \quad (13)$$

The winch shortens the rope if the average position of the moving roller is too low, which means $\bar{x}_{\text{spring,dyn}} < -\varepsilon_{\text{dyn}}$, and lengthens it if the average position is too high ($\bar{x}_{\text{spring,dyn}} > \varepsilon_{\text{dyn}}$). Furthermore, the winch starts operation immediately if the moving roller is approaching one of its stop limits. The calculation of the average position of the moving roller $\bar{x}_{\text{spring,dyn}}$ (13) is extended for operation of the winch. This is done to consider the change of the position of the dynamic drive caused by the lengthening or shortening of the unwrapped rope

$$\bar{x}_{\text{dyn}} = \frac{1}{T_{\text{av}}} \int_{t-T_{\text{av}}}^t (x_{\text{dyn}} - v_{\text{rope}} T_{\text{av}}) dt. \quad (14)$$

D. Safety Measures

A special focus was given to satisfy safety requirements. An emergency stop circuit has been implemented in order to cut off the power of the Lokolift system in case of any error or emergency situation. The emergency stop circuit is activated, if any software plausibility test fails, if the measured unloading force becomes less than 3 kg, if one of the proximity switches (Fig. 2[Q], [R]) is activated, or if the emergency stop button is pressed by the therapist or the subject.

The winch (Fig. 2[E]) is self locking and remains in its actual position in case of power brake down or emergency stop activation. Also, the spring adjustment drive (Fig. 2[H]) is self locking, thus, the elongation of the springs will not change in such a situation. Thus, the subject unloading remains intact with a force equal to the spring force kx_{spring} .

The polyester rope (Fig. 2[A]) has been validated in a test stand before it had been chosen for the Lokolift system. A two year daily use of the unloading system has been simulated with a hydraulic actuator. Thereafter, the rope was torn to test the safety factor of a used rope. As a result, the rope has a sixfold safety factor regarding tear strength. Furthermore, simulation

of daily use of the system was repeated with a shorter simulated time period but with blocked pulleys. Finally, the rope was tested with a 150 kg mass, which dropped several times into the rope. Possible damage of the rope is visible and can be recognized in a regular check. After two years of usage, the rope will be replaced by a new one.

All the points mentioned above are part of a safety analysis of the Lokolift system, that has been a prerequisite for approval of the Lokolift as a medical product in Europe and in the USA.

E. Evaluation of Lokolift

A technical validation of the Lokolift system has been performed with metal plates and a single subject walking on a treadmill. Therefore, iron plates with weights of 53 kg were attached to the end of the rope (Fig. 2[A]). Vertical oscillations of the plates with amplitudes of approximately 10 cm and frequencies of 0.6 and 1 Hz were generated. The mean and the maximum unloading error were calculated from the measured force signal.

In addition, the system performance when using Lokolift with a subject was compared with the performance obtained with a static BWS system Fig. 1(a) and a counterweight BWS system Fig. 1(b). The static system Fig. 1(a) operated with a 3-m elastic polyester rope connected to a mechanical winch. The height of the subject and, thus, his unloading, was adjusted for the desired unloading with the subject standing on the treadmill. Then, the treadmill started and the unloading was measured during walking with fixed rope length. The counterweight system was equipped with the same elastic polyester rope. The desired amount of unloading could be adjusted by attaching iron weights to the distal end of the rope Fig. 1(b).

The accuracy of all three systems was tested with a healthy subject walking on a treadmill at 2 and 3.2 km/h while being unloaded with a desired unloading of 30 and 60 kg. The body weight of the subject was 75 kg. For this test, the actual unloading force at all BWS system was measured with the same force sensor, which was integrated between the horizontal bar carrying the subject's harness and the end of the rope going to the BWS system (Fig. 2[S]). The input force signal for the Lokolift controller was measured with the different force sensor (Fig. 2[N]) mounted to the pulley above the patient (Fig. 2[B]). Thus, the force signal used for control was separately obtained from the force signal for validation. The ground load during walking was recorded with four piezoelectric sensors integrated in the treadmill and a charge amplifier. The signals of the additional force sensor (Fig. 2[S]), as well as the ground reaction load were recorded by a computer.

The system was further tested at the Rehabilitation Center Valens, Switzerland, within regular treadmill training sessions after approval as a medical device. These sessions were conducted in combination with the automated gait orthosis Lokomat (Hocoma AG, Switzerland) [18], which is a robotic device to perform automated and efficient treadmill training. The patients were asked to fill in a questionnaire to assess performance of the Lokolift system. The rating of the therapists are based on personal communications.

In the meantime, the system is in regular clinical use at the University Hospital Balgrist, Zurich, Switzerland with paraplegic and tetraplegic patients.

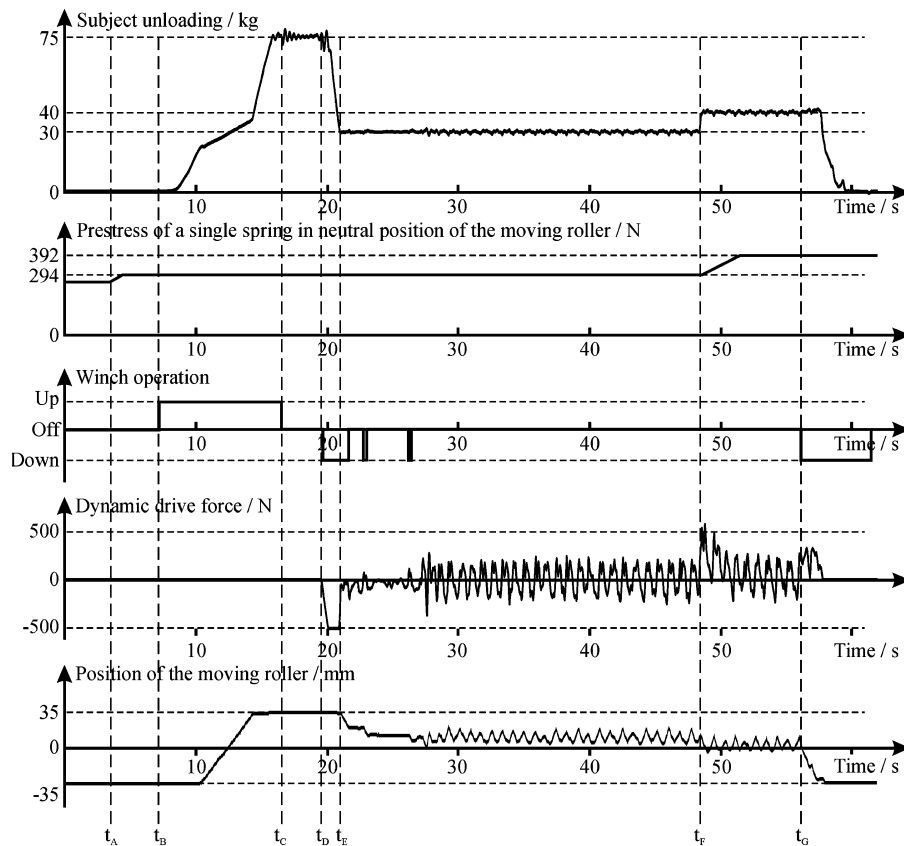


Fig. 7. Sample session with the Lokolift system. A) Desired subject unloading is set and the spring drive starts adjusting the spring length. B) Manual operation of the winch. Subject is lifted up. C) Stop of winch operation. Subject is hanging without ground contact. Subject weight can be assessed. D) Automatic unloading control is started. Subject is automatically moved downwards until the desired unloading is reached. E) Desired unloading is reached for the first time. Dynamic drive starts operation. Winch keeps operating until the dynamic drive is in an optimal working range. F) Change of desired patient unloading. Actual patient unloading is adjusted immediately by the dynamic drive while the spring length adaption takes about 3 s. G) Stop of the dynamic operation and manual downward movement induced by the winch.

III. RESULTS

A. General Statements

The BWS system Lokolift is fully automated and all its functions including lifting of the patient and unloading adjustment can be realized with three computer controlled drives. The system is capable of lifting patients with a body weight of up to 150 kg out of their wheelchair. The maximum dynamic unloading is 80 kg. After activation of the automatic unloading control, the patient gets lifted or lowered until the desired unloading force is reached. Thereafter, the system keeps the unloading force constant. The desired unloading can be adjusted at any time even during the training session without the necessity to stop the treadmill or to lift the patient.

The operation of the Lokolift system and its different drives is demonstrated in Fig. 7. At the beginning, the desired unloading is chosen by the therapist (in this case 30 kg). The spring drive immediately starts operation to adjust the spring length to the desired unloading as described in Section II (Fig. 7, t_A). Thereafter, the therapist starts user-controlled operation of the winch ($t_B - t_C$) so that the subject is lifted out of his/her wheelchair. Finally, dynamic operation of the Lokolift system is activated by the therapist (t_D). The winch moves automatically to reach the desired unloading and to position the dynamic drive in its optimal working range. The dynamic drive starts operation to

keep the unloading force constant as soon as it leaves its top limit (t_E). At a later moment (t_F), subject unloading is exemplarily changed from 30 to 40 kg. The dynamic drive reacts immediately to adjust the unloading whereas the spring length adaptation takes a few seconds.

B. Performance of the Force Controller

When using Lokolift with 53 kg metal plates at oscillation frequencies of 0.6 Hz, the mean error and maximum load error were 1.55 and 2.25 kg, respectively. At a frequency of 1 Hz, the mean and the maximum errors were 2.1 and 3.1 kg, respectively.

At 2 km/h walking speed with 30 kg unloading, the mean and maximum unloading error during dynamic operation of the Lokolift system were 0.44 and 2.37 kg, respectively (Table I). The values increased with higher walking speeds and higher subject unloadings. At 3.2 km/h and 60 kg unloading the mean and the maximum unloading errors were 0.84 and 3.04 kg, respectively. This also indicates sufficiently accuracy of the unloading force measured with the force sensor (Fig. 2[N]) mounted to the pulley above the patient (Fig. 2[B]), because the force error was assessed with the separate force sensor (Fig. 2[S]).

The force is rapidly adjusted by the dynamic drive (Fig. 8). It takes only about 100 ms to adjust the unloading after increasing the desired unloading by 20 kg. Seventy percent of the

TABLE I

UNLOADING FORCE COMPARISONS BETWEEN THE LOKOLIFT SYSTEM, THE STATIC BWS SYSTEM, AND THE COUNTERWEIGHT BWS SYSTEM. THE DATA HAVE BEEN RECORDED WITH A HEALTHY SUBJECT WALKING ON A TREADMILL. THE REFERENCE FOR THE MEAN ERROR, ITS STANDARD DEVIATION (SD) AND THE MAXIMUM ERROR WAS THE MEAN UNLOADING CALCULATED FROM THE RECORDED DATA

		Lokolift	Static system	Counterweight system
Walking speed: 2 km/h Desired unloading: 30 kg	Mean $\pm$ SD	30.0 $\pm$ 0.55 kg	37.1 $\pm$ 6.94 kg	30.1 $\pm$ 1.38 kg
	Mean error	0.44 kg	5.67 kg	1.08 kg
	Max. error	2.37 kg	20.85 kg	5.11 kg
Walking speed: 3.2 km/h Desired unloading: 30 kg	Mean $\pm$ SD	30.0 $\pm$ 0.76 kg	34.5 $\pm$ 10.17 kg	30.1 $\pm$ 3.89 kg
	Mean error	0.61 kg	8.89 kg	3.46 kg
	Max. error	2.38 kg	27.67 kg	8.02 kg
Walking speed: 2 km/h Desired unloading: 60 kg	Mean $\pm$ SD	60.1 $\pm$ 0.88 kg	60.1 $\pm$ 8.02 kg	59.3 $\pm$ 3.59 kg
	Mean error	0.76 kg	8.50 kg	3.01 kg
	Max. error	2.37 kg	19.69 kg	9.66 kg
Walking speed: 3.2 km/h Desired unloading: 60 kg	Mean $\pm$ SD	60.1 $\pm$ 0.99 kg	56.5 $\pm$ 12.5 kg	59.3 $\pm$ 6.15 kg
	Mean error	0.84 kg	11.02 kg	5.34 kg
	Max. error	3.04 kg	26.17 kg	16.22 kg

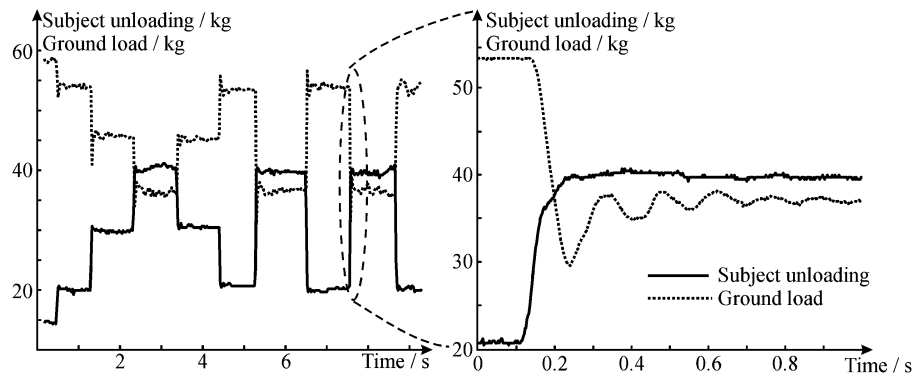


Fig. 8. Step response of the Lokolift force controller with load amplitude of 10 and 20 kg (left). Desired subject force is adjusted within 100 ms at a unloading step of 20 kg (right).

unloading is already adjusted after about 50 ms. The ground reaction force decreased with increasing subject unloading (Fig. 8).

Because of the highly dynamic response of the force controller, it is possible to apply gait cycle dependent force patterns. One example is to perform different unloading forces for the intact and disabled leg in stroke patients (Fig. 9, right). Pronunciation of the stimulation of the afferent receptors during heel strike is also possible by reducing unloading, which, in turn, increases the foot contact force. The unloading drops immediately after the heel strike and increases continuously during stance phase (Fig. 9, left). This pattern may even simulate ground reaction forces when walking down a ramp. The mean force errors (standard deviations) of the saw tooth pattern (Fig. 9, left) and the leg dependent unloading (Fig. 9, right) are 27.7 N (60.3 N) and 10.0 N (13.6 N), respectively.

C. Comparison of the Lokolift System With the Static and Counterweight BWS System

The mean unloading error, its standard deviation, and the maximum unloading error for all three BWS systems are listed in Table I. The largest unloading error was observed at the static BWS system (Fig. 10). At 2 km/h walking speed, the mean and the maximum errors of the counterweight system are more than twice of those of the Lokolift system. This difference increases with higher walking speeds and unloadings. At 3.2 km/h and 60 kg body weight support the errors of the counterweight system

are five to six times larger than the errors of the Lokolift system.

Walking with the static BWS system is characterized by a lower vertical displacement than walking without BWS. When the subject's body is at its lowest point, the fixed rope gets stretched, thus, pulling the subject in an upward direction. In contrast, at the highest point the fixed rope relaxes and the unloading force is reduced. This effect is inverted when using counterweights. The main component of the error force results from inertial effects of the counterweight. Thus, the error force is proportional to the acceleration of the subject's body but oriented in the opposite direction. The subject's body is accelerated upwards when it is located at its lowest point, thus, reducing the subject's unloading and increasing the ground reaction load. Accordingly, the subject's unloading is increased when the body is at the highest point. As a result, the error force enhances the vertical body movement. The subject can even lose ground contact at higher walking speeds and unloading forces as a result of the interfering inertial forces. As the unloading force is constant with the Lokolift system, the vertical movement of a subject is neither inhibited nor enhanced by the BWS system.

D. First Clinical Results

Around 80 regular training sessions have been conducted within the two week trial period at the Rehabilitation Center Valens. The therapists working with the system in Valens and at thereafter at the Balgrist University Hospital gave positive

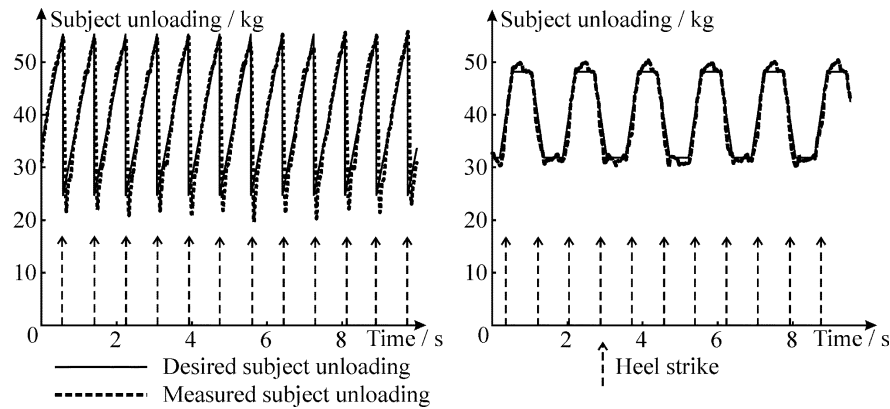


Fig. 9. Highly dynamic gait cycle dependent subject unloading with Lokolift. Subject unloading is reduced at heel strike (left). Different amount of unloading is used for the right and the left foot (right).

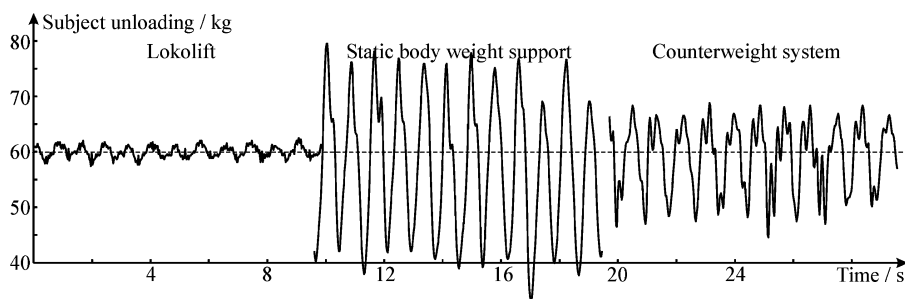


Fig. 10. Measured unloading of the Lokolift system, the static BWS system, and the counterweight BWS system (60 kg unloading weight) during walking of a healthy subject at 3.2 km/h.

feedback. All of them enjoyed the high user comfort and the possibility to adjust the patient weight at any time with an arbitrary amount of unloading. It was appreciated that the patient weight can be assessed just by lifting the patient off the ground and then recording the actual unloading force. The unloading during each training session can be stored automatically for documentation.

Although the system was used for 80 regular training sessions with 11 hemiplegic patients in Valens, only three patients had the cognitive ability required to fill out a questionnaire. All three patients agreed that treadmill training with Lokolift is more convenient than walking with the previously used counterweight system. Two of them said that they can concentrate much better on walking and one answered that he/she can concentrate better on walking. All three indicated that they would like very much to train again with the Lokolift system.

IV. DISCUSSION

A. Technical Statements

The mechanical setup of Lokolift was designed accordingly to the design criteria outlined in Section II. All parts of the dynamic drive unit of Lokolift were made with lightweight materials. The pulleys were made out of plastic and equipped with ball bearings to minimize inertia and friction.

The evaluation of the Lokolift system included the main factors influencing the performance of the system. The variation of stimulation frequency and stimulation amplitude of the control system was considered by increasing walking speed of the subject. Varying operation points were tested exemplarily for two

different unloading settings. Furthermore, the mechanical properties of the coupling between the BWS system and the subject were considered by the tests with weight plates and with a human subject.

The mean load error is greater when using Lokolift with metal plates rather than with a subject. This is due to the mechanical properties of the coupling between the Lokolift system and the load, i.e., metal plates or the subject. The metal plates are directly coupled to the rope of the Lokolift system, whereas the human subject is connected via a harness. Thus, the mechanical properties of the coupling between the subject and the Lokolift system get smoothed due to the soft tissue of the human and slipping effects of the harness.

It is well known that dynamic properties of closed-loop force controlled robotic devices get worse if the coupling between the robotic device and the object, to which the force is applied, gets stiffer [19]. From this point of view, the metal plates connected to the Lokolift system form a worst case consideration, which explains why the mean force error with metal plates is greater than with the subject.

B. Comparison of the Static BWS, Counterweight and Lokolift Systems

It is inconvenient to adjust the mean unloading of the static BWS system if the unloading is not adjusted automatically with a force sensor and an electric winch. A static BWS system cannot guarantee that the unloading force remains constant during the training session. The harness worn by the patient may slip especially at the beginning of a training session. Thus,

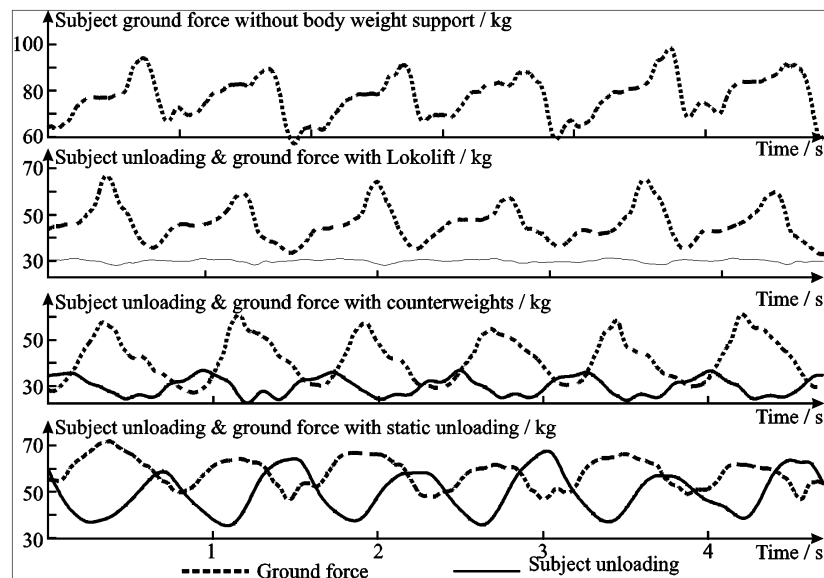


Fig. 11. Ground reaction force and actual unloading during walking without BWS, with the Lokolift system, with the static BWS system, and with counterweights.

the unloading of the patient needs to be adapted after some gait cycles. Furthermore, the unloading force is not constant at all as the maximum unloading error is about 20 and 27 kg at walking speeds of 2 and 3.2 km/h, respectively (Table I).

The counterweight system guarantees a constant unloading as long as the patient does not move. But the inertial forces generated by acceleration and deceleration of the counter mass prevent a constant unloading as soon as the patient starts walking on the treadmill. The higher the walking speed of the patient and the higher the patient unloading, the higher is the oscillation of the unloading force. Furthermore, a change of the patient unloading is cumbersome, because the therapist needs to load the distal end of the rope with a discrete number of metal weights. It is not possible to change the patient unloading continuously, e.g., while the patient is walking.

The Lokolift system produced the smallest errors compared to the static BWS system and the counterweight system. These errors also increase with higher load and higher walking speed, but the mean force error remains at less than 1 kg.

In conclusion, walking with the novel mechatronic Lokolift BWS system can be compared to walking at reduced gravity and constant body mass. Walking with the counterweight system can be interpreted as walking at reduced gravity, but with the mass of the counterweights bagged in a backpack. The main design goal formulated in the introduction section is achieved. The unloading error is clearly lower than that of commonly used BWS system.

The Lokolift system enables highly dynamic force changes. Two examples of gait cycle dependent unloading patterns are given (Fig. 9). To the best knowledge of the authors, no other BWS system has been presented up to now, which allows force variations of up to 20 kg within only 0.1 s.

C. Effects of Oscillations of the Unloading Force

Errors of the desired unloading force yield nonphysiological ground reaction forces, thus, affecting the afferent signals generated by the pressure receptors of the feet. First results show

that walking of a healthy person with the static or counterweight BWS system results in a different ground reaction pattern than walking without BWS (Fig. 11).

The ground force pattern recorded without BWS shows a cyclic plateau phase followed by a ground force peak before the ground force drops to its minimum (Fig. 11). A similar ground force pattern can be observed when walking with Lokolift. In contrast, this ground load pattern could not be observed when walking with the static or counterweight BWS system (Fig. 11). However, physiological gait kinematics and ground reactions with the accompanying physiological afferent feedback are believed to be important for successful gait rehabilitation [20]. In conclusion, the Lokolift BWS system enhances a physiological ground reaction pattern and, thus, has the potential to maximize the therapeutic outcome.

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REFERENCES

- [1] K. E. Gordon, D. P. Ferris, M. Robertson, J. A. Beres, and S. J. Harkema, "The importance of using an appropriate body weight support system in locomotor training," *Soc. Neurosci., Abstract 26* (PN 61.9), 2000.
- [2] H. Barbeau, M. Wainberg, and L. Finch, "Description and application of a system for locomotor rehabilitation," *Med. Biol. Eng. Comput.*, vol. 25, pp. 341–344, 1987.
- [3] K. Norman, A. Pepin, and H. Barbeau, "A treadmill apparatus and harness support for evaluation and rehabilitation of gait," *Arch. Phys. Med. Rehabil.*, vol. 76, pp. 772–778, 1995.
- [4] G. Chen, D. Schwandt, H. F. M. Van der Loos, J. Anderson, D. P. Ferris, F. E. Zajac, S. A. Kautz, C. G. Burgar, C. Patten, R. R. Neptune, and K. E. Gordon, "Compliance-adjustable, force-sensing harness support for studying treadmill training in neurologically impaired subjects," presented at the 6th Annu. Gait Clin. Movement Anal. Meeting, Sacramento, CA, 2001.

- [5] J. He, R. Kram, and T. A. McMahon, "Mechanics of running under simulated low gravity," *J. Appl. Physiol.*, vol. 71, no. 3, pp. 863–870, 1991.
- [6] R. Kram, A. Domingo, and D. P. Ferris, "Effect of reduced gravity on the preferred walk-run transition speed," *J. Exp. Biol.*, pp. 821–826, 1997.
- [7] G. Cook, "Exercise hoist," U.S. Patent US 2002/0065173 A1, 2002.
- [8] F. Gazzani and A. Fadda, "Ward: A pneumatic system for body weight relief in gait rehabilitation," *IEEE Trans. Rehab. Eng.*, vol. 8, no. 4, pp. 506–513, Dec. 2000.
- [9] C.-Y. Lee, K.-H. Seo, C. Oh, and J.-J. Lee, "Newly designed rehabilitation robot system for walking-aid with pneumatic actuator," *Int. J. Human-Friendly Welfare Robot. Syst.*, vol. 4, no. 1, pp. 42–46, 2003.
- [10] S. Kim, H. Shin, S.-H. Jung, J.-J. Lee, and B.-O. Kim, "Supporting force control of walking training robots," *Int. J. Human-Friendly Welfare Robot. Syst.*, vol. 3, no. 3, pp. 2–7, 2002.
- [11] J. Hollerbach, I. Hunter, and J. Ballantyne, *A Comparative Analysis of Actuator Technologies for Robotics*. Cambridge, MA: MIT Press, 1991, pp. 299–342.
- [12] M. Frey, J. Hoogen, R. Burgkart, and R. Riener, "Haptic Interaction with a virtual knee joint—The 9 DOF haptic display of the Munich Knee Joint Simulator," *Presence*, to be published.
- [13] D. A. Laurence, "Impedance control stability properties in common implementations," in *Proc. IEEE Int. Conf. Robot. Automation*, Philadelphia, PA, Apr. 1988, pp. 1185–1190.
- [14] K. E. Gordon, B. Svendsen, S. Harkema, and S. El-Alami, "Closed-loop force controlled body weight support system," World Patent, WO 03/035184 A1, 2003.
- [15] D. Kelsey and A. Walls, "Therapeutic unloading apparatus and method," U.S. Patent US5273502, 1993.
- [16] D. Surdilovic and R. Bernhardt, "String-man: A new wire robot for gait rehabilitation," in *Proc. IEEE Int. Conf. Robot. Automation*, New Orleans, LA, 2004, pp. 2031–2036.
- [17] G. A. Cavagna, N. C. Heglund, and C. R. Taylor, "Mechanical work in terrestrial locomotion: Two basic mechanisms for minimizing energy expenditure," *Am. J. Physiol.*, vol. 233, pp. R243–261, 1977.
- [18] G. Colombo, M. Joerg, R. Schreier, and V. Dietz, "Treadmill training of paraplegic patients using a robotic orthosis," *J. Rehabil. Res. Dev.*, vol. 37, pp. 693–700, 2000.
- [19] R. J. Adams, M. R. Moreyra, and B. Hannaford, "Stability and performance of haptic displays: Theory and experiments," in *Proc. ASME Int. Mech. Eng. Congress Exhibition*, Anaheim, CA, 1998, pp. 227–234.
- [20] V. Dietz, R. Mueller, and G. Colombo, "Locomotor activity in spinal man: Significance of afferent input from joint and load receptors," *Brain*, vol. 125, pp. 2626–2634, 2002.

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