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Body weight support by virtual model control of an impedance controlled exoskeleton (LOPES) for gait training.

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Abstract— **The feasibility of an alternative method to support body weight in a powered exoskeleton is demonstrated. Instead of using an overhead suspension system, body weight is supported by augmenting the joint moments through virtual model control. The advantages of this novel method is that it allows for independent support of the left and right leg, and does not interfere with the excitation of cutanous afferents and balance of the body or trunk. Results show that after a short familiarization period the activity of muscles during initial stance reduces and kinematics become close to normal.**

I. Introduction

Providing patients with body weight support is an efficient way to train neurological patients in producing the required coordinated activity for walking, while they lack the muscle force to counteract gravity. Body weight support is often combined with providing either manual or robotic assistance. Several robotic devices for gait rehabilitation, i.e. Lokomat [1], Gait Trainer [2] , and AutoAmbulator (http://www.autoambulator.com), are available on the market and others have been developed, like the PAM and POGO devices of UC Irvine [3], the LOPES exoskeleton of University of Twente [4]. Most of these devices use a body weight support (BWS) system that consists of an overhead suspension system to unload the body weight. By unloading body weight, patients have to generate less activity to counteract gravity and can utilize the limited remaining muscle force to relearn coordinated stepping movements.

BWS systems are also used in combination with a treadmill without robotic devices where physical therapists support or facilitate the movement of patients. In different cross sectional studies with stroke patients, body weight support has shown to increase the percentage of single limb support time on the paretic limb and to improve temporal symmetry ([5]. Lamontagne and Fung [7] showed that in over ground walking body weight support caused less circumduction and larger hip excursions of the paretic leg. These changes on the short term have a beneficial effect on the long term as different studies showed that providing body weight support during multiple training sessions resulted in improved walking ability [8-11].

The use of an overhead suspension system (OSS) has some disadvantages. An OSS not only provides a vertical force but also a force in the horizontal plane that helps to stabilize the human body or only trunk -in case the pelvis is fixated-. Although in many cases both balance of the body and weight bearing have to be supported, it is a limitation that the amount of support can not be varied independently for both functions. Another limitation in OSS that uses a counterweight or spring mechanisms is that the body weight is supported for both the paretic and non paretic leg, while in many cases it is not needed to support the non paretic leg. To allow independent loading of the legs, mechatronic BWS systems have been developed such as the Lokolift. This system allows variations of the support of up to 20 kg within 0.1 s, [12]. Third, reducing the body load also reduces the excitation of plantar cutaneous afferents. These afferents can adjust motoneuronal excitability, which may contribute significantly to the control of human posture and locomotion.[13, 14].

In this paper we propose an alternative for OSS to support body weight bearing during robot-aided gait training with powered exoskeletons. The presented approach allows for independent support of the left and right leg, does not interfere with the excitation of cutanous afferents and balance of the body or trunk. This is achieved by using a specific implementation of virtual model control [15] (VMC) to modify the exerted torques by the robot. In short, a virtual force is defined that partially counteracts the gravitational force. This virtual force is transformed to joint torque commands for the powered exoskeleton LOPES, which augment the torques of the subject walking in the exoskeleton. The goal of this research is to investigate whether VMC based support of weight bearing is possible in powered impedance controlled exoskeletons and how it affects human gait, especially whether the muscle activity during stance reduces. Preliminary data of kinetics, kinematics, and muscle activations show the feasibility of this new approach.

II. Methods

A. Virtual Model Control

VMC was used to selectively support subtasks of human

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gait. This method allows for straightforward translations between the described interventions and the required joint torques. Using virtual components such as inertias, springs, dampers and force sources it is possible to simulate any interaction that a therapist would usually have with a patient.

To support body weight bearing, a VMC that exerts a vertical force downwards at the ankle is defined:

$$F_z = -S_{BWS} M_{Body} f(X_{rel});$$
$$X_{rel} \geq 0 \Rightarrow f(X_{rel}) = 1; \qquad (1)$$
$$X_{rel} < 0 \Rightarrow f(X_{rel}) = \frac{X_{rel}}{X_{rel}^{\min}}$$

Where S_{BWS} is the fraction of the body weight that have to be supported, M_{Body} is a subject's body weight, X_{rel} is the relative ankle position with respect to the hip, $X_{rel}^{\min}$ is the minimum of the relative ankle position of the previous step, and $f()$ is a scaling factor that is one in case the ankle is in front of the hip and gradually decreases if the ankle is behind the hip to zero at expected toe off condition. The VMC is switched on and off at heel contact and toe off of the corresponding leg, respectively.

The required forces are delivered by knee and hip joint torques transferred, from the robotic exoskeleton to the human endoskeleton. The forces of the virtual components are mapped to joint torques by:

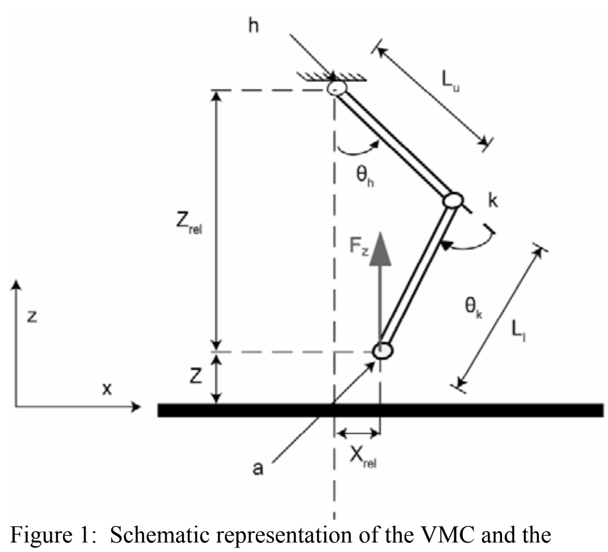

Figure 1: Schematic representation of the VMC and the exoskeleton. With Z the absolute ankle height, L_u the upper leg length, L_l the lower leg length and θ_k and θ_h respectively the knee and hip angles. The virtual force F_z is transformed to joint moments exerted by the exoskeleton to the endoskeleton.

$$\tau = {}_h^a J^T \begin{pmatrix} 0 \\ F_z \end{pmatrix} \qquad (2)$$

Where τ are the joint torques needed to offer the virtual force in Cartesian coordinates (see Fig. 1) and ${}_h^a J^T$ is the transpose of the Jacobian that maps the hip $(\dot{\theta}_h)$ and knee $(\dot{\theta}_k)$ angular velocities to the velocities of the ankle in Cartesian coordinates expressed with respect to the coordinate frame of the hip.

B. Subjects

One young male (24 years, 73 kg, 1.80 m.) volunteered to participate in this experiment. The participant provided informed consent before testing began.

C. Experimental apparatus and recordings

For the experiments the prototype of the gait rehabilitation robot LOPES was used. LOPES is an exoskeleton type rehabilitation robot. It is lightweight and actuated by Bowden cable driven series elastic actuators [17]. The robot is impedance controlled, which implies that the actuators are used as force (torque) sources. The torque for each joint is controlled with an inner loop controller that a force control bandwidth 16 Hz [18].

During all trials muscle activation patterns were recorded by bi-polar surface electromyography (EMG) from the gastrocnemius medialis, tibialis anterior, biceps femoris, vastus lateralis, and gluteus maximus muscles of the right leg. Skin preparation and the placement of the disc-shaped solid-gel Ag/AgCl-electrodes in a bipolar configuration were performed according to Seniam guidelines [19]. For the EMG recordings a compact measurement apparatus (type Porti 16-5, TMS International, Enschede, The Netherlands) was used. The analog signals were sampled at 1024 Hz and sent from the portable unit via fiber optics to the computer, where data were stored for further processing along with the joint angles and torques of LOPES.

D. Protocol

To let the subject become familiar with the device they walked for three minutes in the zero impedance mode [17] with a constant speed of 2.7 km/h. In this mode the impedance of every joint is set to zero, so the robot provides minimal resistance/assistance to the subject walking within LOPES. After a short break the subject walked one minute in the zero impedance mode. Then the body weight support was set to 40% of the body weight for both legs and the subject walked for three minutes. Next the body support was switched off, and the subject walked for one more minute.

E. Data analysis

All EMG processing was done with custom written software [20] in Matlab (Nattick, USA). The raw EMG data were band pass filtered at 10-400 Hz with a 2nd order zero-lag Butterworth filter and converted to smooth rectified EMG signals (SRE) using a low-pass 2nd order zero-lag Butterworth filter at 25 Hz for smoothing. To visually inspect the raw and smooth rectified signals, they were broken up into the individual stride cycles. If one of the muscles contained artifacts (contact artifact, measurement noise) the activity during this cycle was rejected from further analysis.

The joint angles and vertical position of the 'pelvis' and the joint torques exerted by LOPES and the SRE were averaged over five gait cycles and plotted for just before the VMC was switched on, after the VMC was switched on, before the VMC was switched off, and after the VMC was switched off.

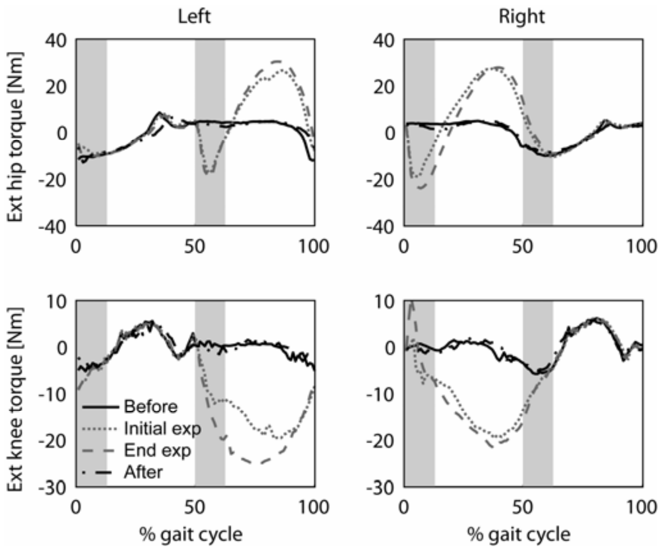

Fig. 2. Joint torques exerted by the robot as function of the percentage gait cycle. The trajectories are averages over five steps immediately preceding exposure to the robot aided weight bearing (before), the first (initial exp) and last (end exp) five steps of robot aided weight bearing and five steps immediately following exposure. The shaded areas indicate the double stance phases. 0 and 100 % indicate the moment of right heel strike. Hip flexion and knee flexion are defined as positive

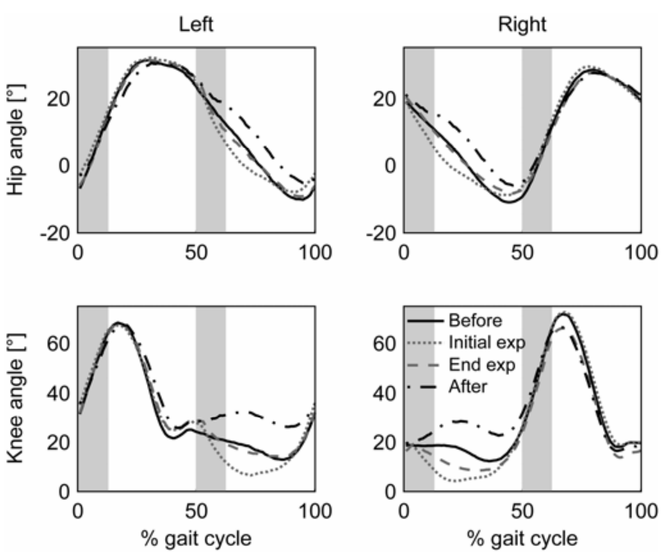

Fig. 3. Average joint angles as a function of percentage gait cycles. See for a further explanation Fig 2.

III. Results

To assist the subject in weight bearing, the robot exerted torques at the knee and hip (Fig. 2). At the knee, an extension torque was generated, whereas at the hip initially an extension torque was generated that changed into a flexion torque as soon as the ankle moved behind the hip. The exerted torques by the robot increased during exposure, especially for the left knee. This indicated that the subject relied more on the robot.

The exerted torques resulted in an exaggerated knee extension and hip extension during initial exposure (Fig. 3). During final exposure the trajectories were returned to normal again. This could be due to an increased reliance of the subject on the robot and a reduction on self generated torques. The observed joint angles when the robot assistance was turned off confirmed this notion. The subject fell short in adequately supporting his weight as demonstrated by an increased knee flexion and hip flexion immediately after exposure.

The reliance of the subject on the external torques was also reflected in a decrease of the major knee extensor, the vastus lateralis during exposure (Fig. 4). Apart of this change, the other observed changes in the muscle activity were increases of the bi-articular gastrocnemius medialis and biceps femoris during mid-stance. The increased activity of these knee flexor muscles is needed to overcome the knee extension moment exerted by the robot in midstance.

IV. Discussion

The goal was to investigate whether body weight support in a powered exoskeleton is feasible by VMC. The impression of the subject was that the introduced method made walking easier. After switching of the weight bearing controller the walking felt heavier than normal and it was clearly observed that he walked with excessive knee bending during stance. The reduction of knee extensor activity at the end of the three minutes exposure was considerable. Even longer exposure might further decrease muscle activity. Reduction in the hip extensors was less clear. Recording other parts of the Gluteus maximus might show larger effects. The increased activity of knee flexors in midstance could be prevented by faster building off the virtual force or by gradually shifting the application point of the force vector from the ankle to the toe. Alternatively this effect can be used to facilitate plantar flexor muscle activity during mid and terminal stance, which we also observed in manual treadmill therapy.

A limitation of this study is that the ankle was not actuated while it is know that ankles in normal stay not flat on the ground but show a typical rollover pattern. Actuation of the ankle might also be beneficial to assist the push off phase. In the LOPES design push-off can be selectively and partially supported applying forward forces at the pelvis. In the future the design of LOPES will be augmented by ankle actuation. Also the force bandwidth we aim to improve to render even more realistic virtual mechanical components.

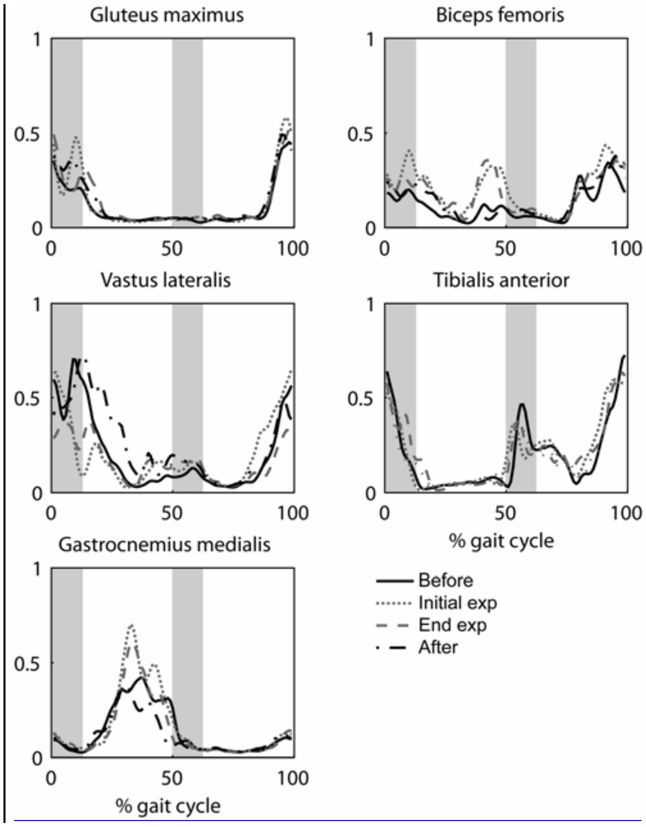

Fig. 4. Average normalized muscle activity as a function of the percentage gait cycle. The muscle activity is normalized to the maximum observed activity in the 20 steps that were used for averaging. See for a further explanation Fig. 2.

V. CONCLUSION

Support of weight bearing in an impedance controlled exoskeleton is feasible and releases the subject from the need to contribute to weight bearing, which was reflected by decreased muscle activity in the loading phase. This novel method is applicable in gait impedance controlled gait trainers and in wearable exoskeletons that augment human muscles.

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