Conference Paper

Voluntary gait speed adaptation for robot-assisted treadmill training

Author(s):
Koenig, Alexander; Binder, Carmen; von Zitzewitz, Joachim; Omlin, Ximena; Bolliger, Marc; Rie ner, Robert

Publication Date:
2009

Permanent Link:
https://doi.org/10.3929/ethz-a-006110521

Rights / License:
In Copyright - Non-Commercial Use Permitted
Voluntary gait speed adaptation for robot-assisted treadmill training

Alexander Koenig, Student Member, IEEE, Carmen Binder, Joachim v. Zitzewitz, Student Member, IEEE, Ximena Omlin, Marc Bolliger, and Robert Riener, Member, IEEE

Abstract—Robot-assisted gait training currently lacks the possibility of the robot to automatically adapt to the patient’s needs and demands (so called “bio-cooperative control strategies”). It is desired to give the patient voluntary control over training parameters such as gait speed or joint trajectories. We implemented a control algorithm for the driven gait orthosis Lokomat that allows severely disabled stroke patients a limited and safe allowance of influence on their gait speed. To exercise gait symmetry, our algorithm can be configured such that only activity in the paretic leg will cause changes in treadmill speed. The algorithm was successfully tested with eight healthy subjects and six stroke patients.

I. INTRODUCTION

Large numbers of repetitions of limb movements were shown to increase success in neurorehabilitation [1], [2]. Robot-assisted gait [4] and arm [5] therapy has the potential to increase the therapy outcome after stroke. Active participation in gait rehabilitation training was shown to increase the therapy outcome [6]. The same is true for the rehabilitation of the upper extremities when stroke patients are forced to use their paretic arm due to constraint induced movement therapy (CIMT) [7]. The robotic gait orthosis Lokomat was developed at the Balgrist University Hospital, Zurich, to enable patients to repetitively exercise their gait pattern over a longer period of time [8]. In the Lokomat, we define active participation as weighted force measurements in hip and knee joints, which provide information to the therapist and the patient on the patient’s level of activity [9], [10].

To maximally challenge and motivate the patient to active participation, current research efforts are directed towards the development of “bio-cooperative” control strategies. These control strategies recognize the movement intention of the patient and adapt the robotic assistance according to the patient’s force contribution to the walking process [11]. Furthermore, virtual environments are used as they have proven to make repetitive rehabilitation exercises more motivating [12].

One example of a patient-cooperative control strategy is the voluntary control of the patient over his or her walking speed. During standard Lokomat training, the patient walks on a treadmill at a predetermined walking speed which can only be changed by the therapist. We developed a walking speed adaptation algorithm in order to provide the patient with a method to voluntarily change the Lokomat speed by his or her walking efforts and not by any manual switching via keyboard or other electronic buttons. Depending on the forces the patient applies to the orthosis, the patient can voluntarily walk faster or slower on the treadmill. In combination with virtual environments, this algorithm enables the patient to actively modulate his walking speed according to the demands of the virtual reality, i.e. active participation is enabled.

The existing treadmill speed adaptation for the Lokomat [3] has several limitations: first, it only works with avery

Manuscript received 31 January, 2009. This work was supported in part by the Swiss National Science Foundation NCCR on Neural Plasticity and Repair under Project P8 Rehabilitation Technology Matrix and in part by the EU Project MIMICS funded by the European Community’s Seventh Framework Program (FP7/2007-2013) under grant agreement n° 215756.

All authors are with the Sensory Motor Systems Lab, ETH Zurich, Institute of Robotics and Intelligent Systems, Department of Mechanical Engineering and Process Engineering, ETH Zurich, Switzerland, and also with the Spinal Cord Injury Center, University Hospital Balgrist, Medical Faculty, University of Zurich.

Phone:+41-44-386 37 39, Address: TAN E 4, Tannenstrasse 1, 8092 Zurich, Switzerland.

Emails:
koenig@mavt.ethz.ch, cbinder@paralab.balgrist.ch, zitzewitz@mavt.ethz.ch, xomlin@paralab.balgrist.ch, mbolliger@paralab.balgrist.ch, rriener@mavt.ethz.ch.
impairments (e.g. stroke or spinal cord injury patients with little to no walking ability). Our approach provides the possibility to influence the treadmill speed of the Lokomat in position control mode by active participation. Speed changes can only occur once per half-step at the end of each swing phase to make the training steadier for patients. To exercise gait symmetry, the algorithm can be configured such that only activity in the paretic leg of a stroke patient leads to changes in treadmill speed.

II. BACKGROUND

Automatic algorithms for treadmill speed adaptation were developed by several researchers to improve locomotion training [13], [14]. These approaches used the position of the subject on the treadmill for adaptation of the walking speed, and therefore cannot be used for rehabilitation training, as the patients are supported by body weight support systems, unable to change their position relative to the treadmill.

Von Zitzewitz et al. [3] were the first to implement a controller that allows the user to adapt his/her gait speed intuitively during robot-assisted gait training. Thereby, the anterior-posterior force component of the sheer forces between the subject’s feet and the ground are indirectly measured by a sensor configuration mounted behind the subject. The anterior-posterior force sensor is thus located in the Lokomat swing door. Details on the sensory setup can be found in [3]. If the anterior-posterior force points in walking direction, the treadmill accelerates; if directed against the walking direction, the treadmill decelerates. For this approach, the subject needs to walk in zero impedance mode in the Lokomat. In this mode, he/she is able to apply sheer forces on the ground in a voluntary way while the joint torques induced by the orthosis are reduced to a minimum. Although this approach can be used with subjects that have good trunk control and that need little supportive force during Lokomat training, it cannot be used with weaker subjects that need full support force from the Lokomat. In Lokomat position control mode, i.e. at very large supportive force, the controller that guides the orthosis and the patient’s legs on the treadmill is too stiff. Consequently, the controller does not allow deviations from the desired trajectory. This prevents the patient from using his or her stance leg to apply voluntary anterior-posterior forces to accelerate or decelerate. Furthermore, this approach is not appropriate for stroke patients, as hemiplegia will cause an asymmetric gait. Patients would then accelerate more strongly with their healthy leg than with their paretic leg, and the treadmill training would reinforce this asymmetry.

Although the implementation of von Zitzewitz’s approach in the Lokomat path controller [15] provides the possibility of treadmill speed adaptation for patients with limited walking functionality or hemiplegia, it is not suitable for patients with severe neurological disabilities though.

Our algorithm bridges this gap giving severely affected patients with little to no walking capability the possibility to control walking speed in the Lokomat while still providing symmetric walking speed for both legs.

III. METHODS

Instead of anterior-posterior forces that result from the sheer forces onto the ground (Fig. 2) [3], swing leg forces were used to accelerate or decelerate the treadmill speed. In addition to the possibility to work at high levels of support force, this approach has the benefit, that the speed adaptation algorithm can be used in any commercial Lokomat system equipped with force sensors, without having to add force sensors for measurement of anterior-posterior forces.

![Fig. 2. Acceleration while walking in the Lokomat. A: At low guidance forces. The subject creates a torque $T_1$ in the stance leg, which creates a reaction force $F_{\text{sheer}}$ on the treadmill. A proportional anterior/posterior force $F_{\text{A/P}}$ can be measured by the force sensor $S_F$. As anterior/posterior excursions are not possible in the Lokomat, the exoskeleton must be compliant enough to allow deviations from the reference position in the swing leg. This approach has been derived in [3] B: In position control mode. The subject creates the action torque $T_a$ in the swing leg by exerting force $F_a$ onto the orthosis. A reaction torque $T_r$ in the stance leg causes a similar reaction force $F_{\text{sheer}}$ in the stance leg as in figure A. The force sensor $F_s$ is not needed.](image)
When evaluating the new algorithm, anterior-posterior forces were additionally recorded, in order to investigate, whether or not our approach of using the joint forces of the swing leg leads to comparable results as using the anterior/posterior forces (Fig. 2A).

A. Algorithm

For safety reasons, we do not give the subject full control over the treadmill speed but let him or her choose between three speeds: slow, normal and fast. The decision at which speed to walk is based on the estimation whether the patient has the intention to slow down, keep the speed or increase the speed. For security reasons, the acceleration was limited to \(0.07 \text{m/s}^2\) *0.25km/h per second). A base treadmill speed \(v_{\text{base}}\) is set by the therapist and multiplied with a speed multiplication factor (SMF). The real treadmill speed \(v_{\text{real}}\) is then

\[
v_{\text{real}} = v_{\text{base}} \cdot \text{SMF}
\]  

(1)

![Table 1: Description of patients](image)

The aphasic patients were not able to answer questions, but able to understand instructions. Subjects 3 and 6 were wheelchair bound. Subjects 1 and 2 were able to walk, but no 10m walking data was available.

We compute the value of SMF from weighted force measurements of hip and knee joints, the so-called biofeedback values, which reflect patient activity and participation during Lokomat training and have been developed in previous studies [9], [10]. These biofeedback values are unitless. The weighting functions were defined for each part of the gait cycle, such that the resulting biofeedback values increase for therapeutically desirable movements, e.g. active knee flexion for early swing.

Calibration is necessary at the beginning of each training session, as the absolute biofeedback values of one subject are different for each Lokomat training session due to slightly altered attachment of the orthosis to the patient legs. We instruct the patient to maximally accelerate or decelerate and record the biofeedback values during gait. The maximum biofeedback value \(B_{f_{\text{max}}}\) and the minimum biofeedback values \(B_{f_{\text{min}}}\) recorded are the values which correspond to strong acceleration or strong deceleration. (Fig. 3A). Then, the therapist defines variable lower and an upper biofeedback boundaries \(B_{f_{\text{low}}}\) and \(B_{f_{\text{hi}}}\) based on the calibration values.

If the current biofeedback is smaller than \(B_{f_{\text{low}}}\), the subject will walk slower, if it is larger than \(B_{f_{\text{hi}}}\), the subject will walk faster, and if \(B_{f}\) is in between \(B_{f_{\text{low}}}\) and \(B_{f_{\text{hi}}}\), the subject will stay at \(v_{\text{set}}\).

B. Study design

To prove the functionality of our approach, we performed two studies: one with healthy subjects and one with clinical patients. Studies were conducted at the Spinal Cord Injury Center of Balgrist University Hospital, Zurich. Healthy subjects were eligible for the study if they had no neurological impairment and had never walked in the Lokomat. Stroke patients were eligible if they had suffered from stroke, had no to little walking capabilities and were able to follow verbal instructions. Osteoporosis led to exclusion from the study. Femur length had to be between

![Fig. 3. Example of a biofeedback calibration and the threshold determination during gait. A: Calibration phase: Calibration is needed, as the biofeedback values differ among patients as well as between trainings. The patient is instructed to walk at normal walking speed (1). Then we instruct the patient to decelerate (2) and accelerate (3). The actual walking speed does not change during calibration. B: Application of the calibrated algorithm during training (without instructions to the patient): The minimum and maximum biofeedback values from the calibration phase are used as a reference to set the thresholds \(B_{f_{\text{low}}}\) and \(B_{f_{\text{hi}}}\). If the biofeedback is larger than \(B_{f_{\text{hi}}}\), the treadmill speed will increase; if it is lower than \(B_{f_{\text{low}}}\), the
35 and 47 cm. Approval for all studies was obtained from local ethics committees, and all subjects gave written informed consent.

Subjects walked for seven minutes to become acquainted to walking in the Lokomat. None of the subjects had experienced Lokomat training before. The subjects were instructed by the experimenter to adapt their walking speed to four predefined speed patterns (Fig. 4). The four patterns were chosen such that any transition between slow, normal and fast walking could be evaluated. The assignment was randomized. We recorded $v_{\text{real}}$, BF$_{\text{min}}$, BF$_{\text{max}}$, BF$_{\text{low}}$, BF$_{\text{hi}}$, the average biofeedback value of hip and knee joints and our instructions to change walking speed. We asked the subjects to fill out a questionnaire after the Lokomat training. In a study with health subjects, we recorded data from eight healthy subjects (4 female, 4 male, mean age 21.3±1.3y). Six stroke patients had been recruited in order to evaluate the speed adaptation algorithm (Table I).

C. Evaluation statistics

When walking over ground at 0.99m/s, humans with intact gait deviate from their mean velocity between -10% and +20% [16]. For the qualitative evaluation of our approach, we computed the amount of time, when the real treadmill speed $v_{\text{real}}$ was within these limits around the desired walking speed $v_{\text{command}}$. To quantify the quality of our algorithm, we computed the percentage of the total training time, during which $0.9v_{\text{command}} \leq v_{\text{real}} \leq 1.2v_{\text{command}}$. We started comparing actual treadmill speed with the treadmill speed commanded by the therapist (Fig. 4) one second after the command was given, as the speed increase was rate limited due to patient safety reasons. For each subject, we computed the mean values for each condition (slow, normal, fast) (Fig. 6).

IV. RESULTS

A. Speed adaptation algorithm

All healthy subjects and patients were able to follow the desired speed pattern within a range of [-10, +20] percent, as exemplarily shown in Fig. 5. The mean over the percentile values for each subject showed that subjects were able to use the algorithm equally well in each condition (slow, normal, fast speed) (Fig. 6). We verified that subjects performed equally in each speed pattern A-D of Fig. 4 (result not shown).

Biofeedback values recorded during the training sessions could not be compared directly, as we needed to re-adjust the Lokomat parameter settings between recordings, leading to biofeedback values of different magnitude for the same level of activity [12].

To quantify the reactivity of the algorithm, we computed the time delay between the command to the subject and the time when the treadmill speed reached the desired speed. From this, we subtracted the time delay caused by the rate limited treadmill acceleration (Fig. 7). On average, the algorithm reacted within less than one half step for healthy subjects (0.5s±3.0) and less than a step in patients (1.3s±2.9).
B. Comparison with existing approaches

To compare our approach to the one of von Zitzewitz [3], we recorded the anterior-posterior forces for one patient and computed the estimated walking velocity based on this force measurement according to [3]. Consequently, we computed the slope, i.e. the first derivative of a linear fit of this computed velocity for each condition (slow, normal or fast). We thereby obtained the theoretical, average acceleration, \( a_i \), for each walking condition. For the evaluation, \( a_i \) was shifted and normalized using:

\[
\bar{a}_i = \frac{\text{mean}(a_i) - \text{mean}(\sum_{i=1,2,3} a_i)}{\text{mean}(\sum_{i=1,2,3} a_i)}
\]

(2)

With \( i \) being the instructed condition: 1=walk slow, 2=keep speed at normal level and 3=walk fast, as in Fig. 4.

The value \( \bar{a}_i \) reflects the patient’s intention to increase his walking speed (\( \bar{a}_i > 0 \)) or to slow down (\( \bar{a}_i < 0 \)). The shift in (2) was necessary, as the patient’s overall mean acceleration \( \text{mean}(\sum_{i=1,2,3} a_i) \) was computed to be negative (-7.92 m/s\(^2\)). This could result, first, from the patient’s incapacity to produce a sufficient amount of positive anterior-posterior force and, second, from a shift of this force component to negative values due to the foot trajectory prescribed in position control mode. The comparison of the output of both algorithms shows that walking in the Lokomat with our algorithm resulted in reasonable gait accelerations (Fig. 8).

C. Questionnaires

We asked each healthy subject and each patient to fill out a questionnaire. The answers are summarized in Table II. Two patients were aphasic and could not answer all questions of the questionnaire. Their answers were excluded.

The healthy subjects rated the speed adaptation as less motivating than the patients, which was to be expected as they possess normal walking function. All patients

<table>
<thead>
<tr>
<th>TABLE II: RESULTS OF THE QUESTIONNAIRES.</th>
<th>n=8</th>
<th>n=6</th>
</tr>
</thead>
<tbody>
<tr>
<td>Speed increased when I wanted to accelerate</td>
<td>6.6</td>
<td>7.1</td>
</tr>
<tr>
<td>Speed decreased when I wanted to decelerate</td>
<td>8.0</td>
<td>7.0</td>
</tr>
<tr>
<td>I could hold the desired speed</td>
<td>7.2</td>
<td>7.1</td>
</tr>
<tr>
<td>The influence of the walking speed motivated me</td>
<td>6.0</td>
<td>9</td>
</tr>
<tr>
<td>I found the increase in speed as a reward for my increased participation</td>
<td>6.2</td>
<td>8.1</td>
</tr>
<tr>
<td>I found the Lokomat training to unsteady</td>
<td>No (7/8)</td>
<td>No(3/4)</td>
</tr>
</tbody>
</table>
reported to feel in control of the training and were motivated by the new possibility to have influence on the training.

V. DISCUSSION AND OUTLOOK

We introduced a new approach for adaptive treadmill training in the Lokomat explicitly developed for weak patients that require a high level of support force from the Lokomat. It was usable by all healthy subjects and patients and reacted for all subjects and patients in average within less than one step. The speed change was rate limited, in order to avoid hasty changes in the training that might make patients feel uncomfortable. The healthy subjects rated the speed increase and decrease of 0.07 m/s² as too slow. In contrast, patients did not judge the reaction time as too slow but rather reported that the speed increase motivated them to participate actively in the training.

The result from Fig. 8 shows that the subject was able to produce an acceleration pattern with consistent and clearly distinguishable values for acceleration, deceleration and while maintaining the desired walking speed. This means that our algorithm did indeed produce similar results as the approach of von Zitzewitz [3]. However, the mean acceleration of (-7.92 m/s²) shows that the approach of von Zitzewitz [3] is not applicable for our patient group. The patients constantly decelerated, even when their intention was to accelerate.

Although no larger clinical trials are scheduled, we are planning to integrate the speed adaptation algorithm into a virtual soccer scenario. In such a virtual environment, we will give the user the possibility to adapt the walking speed of his virtual character in the virtual environment. The patient can e.g. accelerate his/her speed to more quickly approach and catch a soccer ball than a virtual opponent.

ACKNOWLEDGEMENTS

We thank Lars Lünenburger and Markus Wirz for help with the study protocol and the single case studies. We would also like to acknowledge the cooperation with the Zentrum für Ambulante Rehabilitation, Zurich for help with patient recruitment.

REFERENCES


