Feasibility of selective robotic support of foot clearance with continuously adapting impedance levels

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Abstract— **Encouraging stroke survivors to actively participate in robot aided gait training is critical for optimizing the outcome of this intervention. In this respect, it is of crucial importance that the timing of the provided assistance and the amount of assistance is in accordance with the subjects needs. We tested the feasibility of a control algorithm for a powered exoskeleton that selectively supports foot clearance and adapts its support to the performance of the subject. This was done in five chronic stroke survivors with stiff knee gait. Foot clearance was selectively supported through a virtual spring between the desired and actual ankle height of the paretic leg. The virtual spring stiffness was automatically adapted based on the experienced movement error in the previous step and a forgetting factor. The results showed that the virtual spring stiffness profile converged to a steady state pattern in about 20 steps. The pattern was subject specific and was roughly shaped to the deviation of the actual ankle trajectory from the reference trajectory before the assistance was turned on. The assistance resulted in an increased foot clearance through increased knee flexion, whereas it left other aspects of gait unaffected. The presented algorithm turned out to be effective in providing appropriately timed assistance according to the subjects needs.**

Keywords— **Adaptive control, stroke, rehabilitation robots, gait training.**

I. Introduction

Robot aided treadmill training is an emerging intervention to regain walking ability in stroke survivors. Robot aided training allows task specific and intensive training while it does not place a high physical demand on the therapists like in manually assisted treadmill training. As it is a relatively new technology, a lot of research effort is put in improving the design and control of the robotic devices. In first instance, the devices were used to move the legs of the subjects through prescribed patterns by using position control. The major limitations of this approach were that the subjects were not required to actively contribute to the movements and that they did not experience any movement errors. Active participation and making errors have been shown to be crucial in motor learning.

To increase the active participation of the subjects and encourage them to self generate activity, different research groups have opted to utilize interactive control schemes that control the interaction forces between the robot and patient. These interactive schemes open many possibilities but also introduce new challenges, like determining the appropriate level and timing of the provided assistance for each specific subject.

A possible solution for the appropriate timing is to split up gait in different subtasks that need to be accomplished during specific phases of the gait cycle and to selectively control these subtasks during the corresponding phase. For instance, subjects need to accomplish appropriate foot clearance during the swing phase of walking. This can be achieved by controlling the vertical ankle height during the pre swing and the swing phase. By using a gait event detection algorithm, the support can be switched on and off with the appropriate timing. This assures that the control of the support is always in synchrony with the gait events of the subject. We recently showed that using this approach enables us to selectively influence the step height and length of healthy subjects walking in the gait trainer LOPES [1].

A possible solution for determining the appropriate level of support was put forward by Emken and colleagues [2]. They proposed an automated method to adjust the amount of support to the capabilities of the subject, a so called “assist-as-needed” algorithm. This algorithm consists of an error-based learning law with a forgetting factor. It adapts the amount of support on a step-by-step basis such that it reduces the assistance when errors are small and increases its assistance when errors are large, effectively bounding these errors. They showed, in a group of spinal cord injury subjects, that the amount of support was adapted in a subject specific way such that the robot only provided support in those regions of the step cycle where the subject consistently made large errors.

In this study we combined the above described selective subtask control and the adaptive control algorithm. The purpose of the study was to demonstrate the feasibility of this combined algorithm in selectively supporting the foot clearance and adjusting the support to the requirements of the subject.

O. Dössel and W.C. Schlegel (Eds.): WC 2009, IFMBE Proceedings 25/IX, pp. 92–95, 2009.
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II. METHODS

A. Subjects

Five male chronic stroke survivors (age: 57.2±4.2 years, length: 1.82±0.04, weight: 89.2±11.7) volunteered to participate in this experiment. All stroke survivors had a left hemi paresis secondary to a single and first ever unilateral stroke. They were at least half a year post-stroke and were all independent ambulators. Their walking pattern was characterized as stiff knee gait.

B. Experimental apparatus and recordings

For the experiments the prototype of the gait rehabilitation robot LOPES was used. LOPES is an exoskeleton type rehabilitation robot with 8 actuated Degrees of Freedom: pelvic horizontal translations and hip abduction/adduction, hip flexion/extension and knee flexion /extension of both legs. It is lightweight and actuated by Bowden cable driven series elastic actuators [3]. The robot is impedance controlled, which implies that the actuators are used as force (torque) sources.

Applied torques and measured joint angles of LOPES are sampled at 100 Hz. The joint angles and segment lengths are used to calculate the ankle position at each instant of time. Four force sensors integrated in the treadmill measured the vertical forces and were used to calculate the centre of pressure (CoP). The CoP velocity and position were analyzed in real time to detect heel contact (HC) and toe off (TO) events. HC and TO were used as the triggers to switch on and off the robotic support.

The forces, torques and joint angles were stored for offline analysis. From the data we determined different gait parameters and kinematic metrics for each individual step.

C. Protocol

Subjects were strapped into LOPES, such that the hip and knee rotation centers of the exoskeleton lined up with those of the subjects. After being acclimated to walking in the robot, the preferred walking velocity was assessed by systematically varying the treadmill velocity. During these trials the robot was operated in zero impedance mode, meaning that the robot did not apply any support torques. Subsequently, the subjects first walked at this self selected speed for about 1 minute, after which the robotic support was turned on and the subjects continued walking for several minutes. The provided support was solely aimed at increasing the foot clearance (that is the vertical ankle position) of the paretic leg. The support was switched on at heel contact of the non paretic leg and switched off at heel contact of the paretic leg. We defined a reference vertical ankle trajectory, based on a parameterized reference trajectory obtained from measurements with healthy subjects. This pattern was scaled for each subject such that the maximum of this pattern was 5 cm higher than the measured maximal ankle position during walking without support.

D. Adaptive foot clearance support

In controlling the robot we adopted an approach in which the high level control is at a (sub)task level instead of the generally used control at the level of the joints. The used controller is based on the Virtual Model Control (VMC) framework [13]. The basis of this control method is to define physical interactions with the subject that would aid the patient with the concerned gait task. In this case, we define a virtual spring between the actual ankle height and the reference ankle height. If the actual ankle height (Z) deviates from the reference ankle height (Z_{ref}) a "virtual force" is exerted at the ankle which mimics a therapist pulling up the foot at the ankle. The exerted force is dependent on 1) the current deviation of the reference pattern that is movement error, 2) the virtual spring stiffness and 3) on the performance during preceding steps through step to step adaptation of the stiffness of the virtual spring.

In mathematical terms:

$$F_z = K_z(Z - Z_{ref}) \tag{1}$$

where F_z is the desired virtual force and K_z the virtual spring stiffness. The desired forces are delivered by knee and hip joint torques, transferred from the robotic exoskeleton to the human endoskeleton. The virtual force is projected to joint torques by using the Jacobian of the kinematic structure.

The stiffness of the virtual spring was adapted from cycle to cycle according to:

$$K_z^i(t) = f * K_z^{i-1}(t) + g * \left(Z_{ref}^{i-1}(t) - Z^{i-1}(t)\right) \tag{2}$$

Where the subscript i denotes the ith step cycle, f is a forgetting factor set to 0.9, g is an error based gain set to 1800 and t indicates the % gait cycle. The stiffness was constrained to positive values. This scheme to adapt the stiffness is adopted from [2]. The idea behind the scheme is that when the desired ankle height is attained by the subject the support will be reduced, since the forgetting factor (always smaller than one) will reduce the stiffness of the virtual spring for each subsequent step. If the subject is unable to follow the desired ankle height trajectory the virtual spring stiffness will be increased in the next step cycle, proportional to the tracking error and weighed by the error based gain.

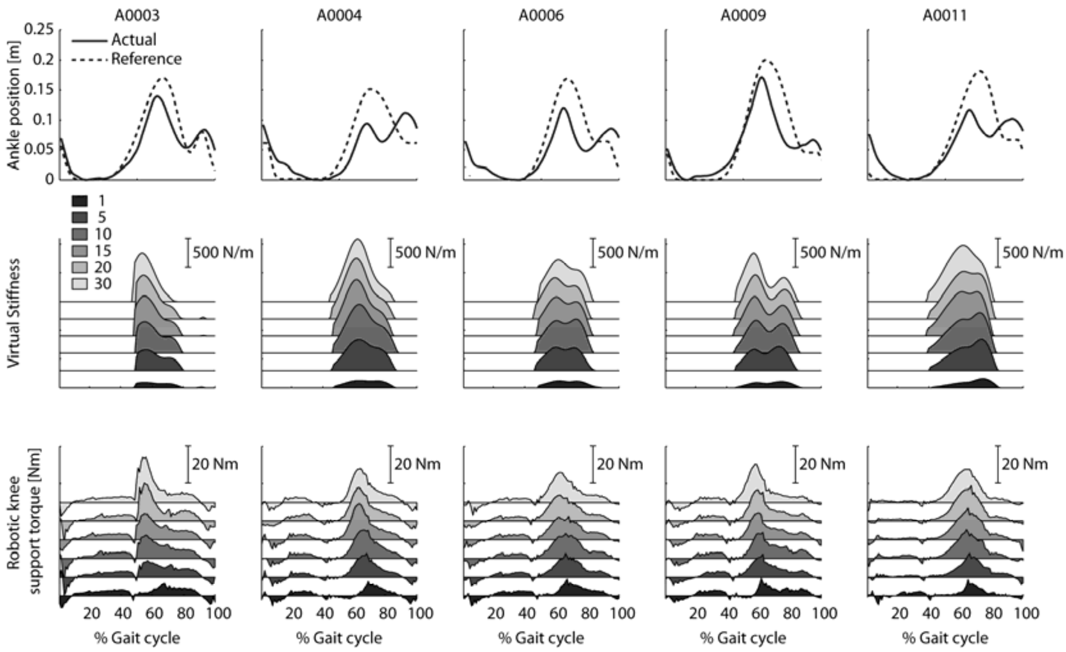

Fig. 1. Shaping of the impedance during the swing phase based on the difference between the actual and the reference ankle trajectory for five chronic stroke survivors. Upper row of graphs indicate the reference and actual ankle trajectory for a step in which no support was provided. The difference between the actual and the reference trajectory is indicative for where robotic support is needed. The middle row of graphs indicates the adaptive virtual stiffness during selected steps (at 1, 5, 10, 15, 20 and 30) after the robotic support was turned on. The virtual stiffness is adapted in every step, based on the error in the previous step and a "forgetting" factor. The lower row of graphs shows the exerted robotic support at the knee. The knee and hip torque (not depicted) are calculated from the virtual stiffness and the deviation between the actual and reference ankle trajectory using a Jocobian transformation.

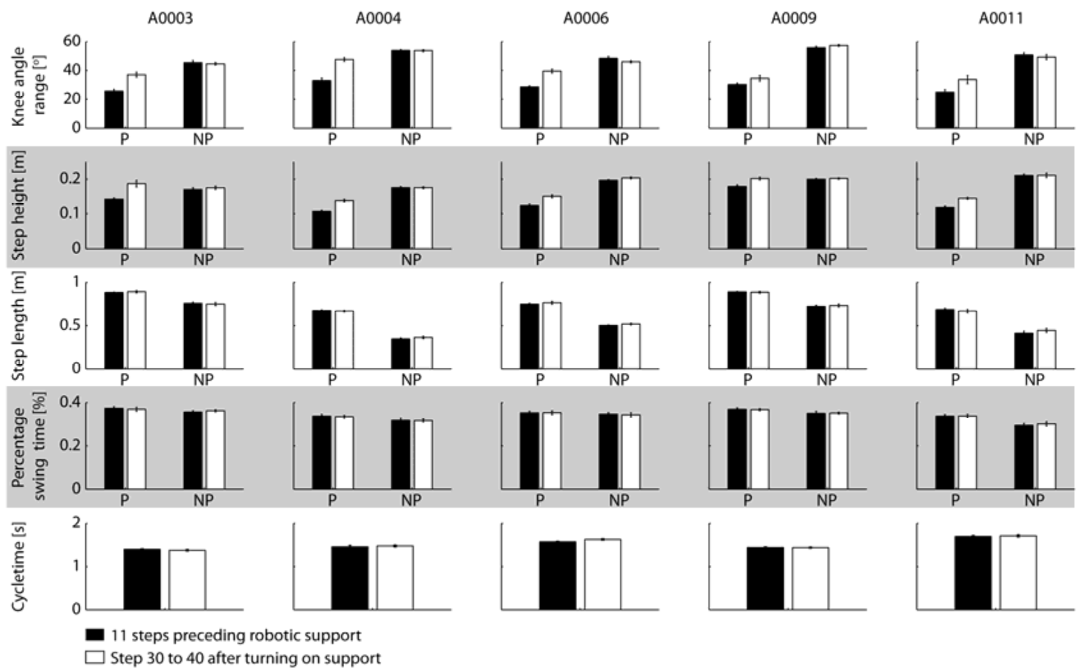

Fig. 2. Assorted kinematic metrics and gait parameters for the different stroke survivors for their paretic (P) and non paretic (NP) leg. The bars represent averages over 11 steps preceding the exposure to robotic support (black) and after the robotic support converged to a "stable" pattern (light grey).

III. Results

The ankle height trajectory of the stroke survivors all deviated in a different way from the scaled reference trajectory (Fig. 1). However, they had in common that the actual ankle height was below the reference ankle height in most of the swing phase (starting from 60%) except the terminal swing phase (>90%). The virtual stiffness gradually increased during the first steps of exposure to the robotic support. Yet, it only increased in the part of the gait cycle where the ankle position was below the reference trajectory, so where the support was needed. In general, the virtual stiffness converged to a steady state pattern in approximately 20 steps. The "steady state" pattern of the virtual stiffness reflected roughly the pattern of the initial deviation between the reference and actual vertical ankle position. The changes in the virtual stiffness were reflected in the applied robotic support at the knee and at the hip joint (not depicted).

The applied robotic support resulted in a significant increase of the knee flexion during swing (see Fig. 2) for all subjects (separate paired t-test for every subject to assess whether the parameter values from the eleven steps prior to applying robotic support differed from those of the eleven steps when a steady state was achieved). The increase in knee flexion resulted in a significant increase of the maximal ankle height during swing. The other parameters were fairly unaffected. Although some parameters showed a significant change in some of the subjects, the relative changes were generally smaller than 3%.

IV. Discussion

In this study we showed the feasibility of combining a control algorithm that specifically supports subtasks of walking with an algorithm that automatically adapts its support to the performance of the subject. The combination of these algorithms resulted in a robot control algorithm that provided support on that part of the swing phase were support was required while it left the other parts of the swing phase and the remaining walking pattern unaffected. The reported results are in agreement with those of Emken and colleagues [2]. They combined an adaptive algorithm with a manual teach and replay algorithm and supported the complete step (step height and step length). Their results, obtained in spinal cord injury subjects, also showed that the stiffness was adapted in a subject specific way. Furthermore they also showed that the convergence to a steady state pattern took about 20 steps.

In the current study we focused on the "convergence" property of the adaptive algorithm. Yet, another advantage of the algorithm is that it keeps adapting the stiffness every step after this initialization phase. In doing so it allows variability in performance and the occurrence of errors. These properties are essential to promote motor (re) learning.

We assessed the instantaneous effects of providing robotic support in a single session. We are currently conducting an effect study in which the presented algorithm is applied in a six week training program aimed at improving the foot clearance in chronic stroke patients during overground walking. Although in a single session the algorithm selectively influence the foot clearance, on the long term the support might also result in beneficial changes of other aspects of the walking pattern, like the asymmetry in step length. Like the majority of stroke survivors, all subjects exhibited a clear asymmetric walking pattern while walking overground. This pattern was also observed while walking in LOPES. This was reflected in the difference between the paretic and non paretic step length. The non paretic foot was placed less far forward with respect to the paretic foot than vice versa (Fig. 2). Difficulties in attaining enough foot clearance are widely regarded as a reason for this asymmetry in step length. Improvements in the foot clearance as a result of the robotic support could therefore also result in less selective effects on the long run.

Acknowledgment

This study was supported by a grant of the Royal Dutch Society for Physical Therapy (KNGF). We want to thank Joelle van den Hoek and Marjanne Folkersma for their help in planning and running the experiments.

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