Low-level finite state control of knee joint in
paraplegic standing

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ABSTRACT

Low-level finite state (locked-unlocked) control is compared with open-loop stimulation of the knee extensor muscles in functional electrical stimulation (FES) induced paraplegic standing. The parameters were: duration of standing, relative torque loss in knee extensor muscles, knee angle stability, average stimulus output and average arm effort during standing. To investigate the impact of external mechanical conditions on controller performance, experiments were performed both under the condition of a freely moving ankle joint and of a mechanically stabilized ankle joint. Finite state control resulted in a 2.5 to 12 times increase of standing duration or in a 1.5 to 5 times decrease of relative torque loss in comparison with open-loop stimulation. Finite state control induced a limit cycle oscillation in the knee joint. Average maximum knee flexion was 6.2° without ankle bracing, and half that value with ankle bracing. Average arm support was 13.9 and 7.5% of the body weight without and with ankle bracing respectively.

Keywords: Finite state control, paraplegic standing, knee joint, ankle bracing

INTRODUCTION

Functional electrical stimulation (FES) has been demonstrated to be feasible in assisting paraplegic individuals to stand up, to remain standing and to walk; the clinical possibilities of such systems are at present limited. One major reason for this is the undue stressing of the stimulated muscles in the current open-loop systems, leading to early fatigue. However, during standing the vertical body orientation in principle allows a reduction in the activation of muscles involved, as the mechanical structure of the patient then supports most of the effort. As the body in the upright position is critically stable, feedback control of muscle activation is required to give the patient the physical stability, and the confidence, to be functional. The feasibility of finite state closed-loop control for this purpose has been demonstrated. Robust control of the knee joint was achieved in experiments on paraplegic subjects who were in the supine position and the resulting dynamic muscle activation was shown to reduce fatigue in knee extensor muscles.

The actual benefit of closed-loop control with respect to postponing fatigue in knee extensor muscles depends on the load situation at the knee joint, and will be maximal when no extending joint moment is required. During standing, under the condition of minimal upper body loading, the external load situation at the knee is determined by the position and the amplitude of the ground reaction force (GRF) and the subject's posture. Important in this is the point of application of the ground reaction force; the position of this point relative to the body depends on the ankle joint moment. As the latter may be influenced by (passive or active) ankle stabilization this offers the possibility for indirect stabilization of the knee joint without knee extensor activation.

The current study was initiated to evaluate the applicability of finite state artificial reflex control for the stabilization of the paraplegic knee joint under actual standing conditions. To evaluate the controller in two important daily life and clinical approaches, experiments were performed under the condition of both a freely moving and of a mechanically stabilized ankle joint. The present study compares finite state closed-loop control with traditional open-loop stimulation both with respect to knee joint stability and fatigue occurring in the activated muscles.

BIOMECHANICAL ANALYSIS

We consider the typical approach of standing by FES, in which the knee joints are stabilized in extension by activation of the quadriceps muscles and the hip joint is stabilized near the neutral position either by stimulation of hip extensor muscles or by mechanical bracing (Figure 7). If the model is considered to be static, and the contribution of the upper limbs (which are used for balancing, as the model is critically stable) that has to be developed by the knee joint muscles for
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Figure 1 The biomechanical model of standing. GRF is the ground reaction force, \( l \) the length of the lower leg, \( M_b \) the trunk mass, \( \theta \) the angle of the lower leg relative to the vertical, \( L \) the length of the part of the foot anterior to the ankle joint and \( p \) the point of application of GRF. Hip and trunk are stabilized by the upper part of a reciprocating gait orthosis. Indicated is a situation with flexing knee moment and (passive) ankle moment as GRF passes behind the knee and in front of the ankle joint axis. When \( T_k \) is the ankle moment the horizontal position of the point of application \( p \) of GRF relative to the ankle joint is given by:

\[
T_k = \frac{1}{2} M_b g l \sin (\theta) - T_k, \text{pas}\]

\(\text{for each leg can be written as:}\)

\[
T_k = \frac{1}{2} M_b g l \sin (\theta - T_k, \text{pas}) + T_k, \text{pas}
\]

where \( T_k, \text{pas} \) and \( T_a \) denote the passive moments at knee and ankle joint respectively. The term \( T_k, \text{pas} / \left( \frac{1}{2} M_b g l \right) \) equals the position of the GRF relative to the ankle joint axis. \( T_k, \text{pas} \) and \( T_a \) may result from ligaments, contracture or increased tone of flexor muscles, or mechanical bracing and are in general nonlinearly related to the respective joint angles. Positive values of \( T_k \) and \( T_a \) correspond to flexing knee moment and plantar flexing ankle moment respectively.

Under static standing conditions the moment balance at the knee joint must be such that the quadriceps torque is equal to \( T_k \) to stabilize the knee. Under the assumption of neglectable passive knee and ankle moments, this would mean that a posture resulting in e.g. \( \theta = 5^\circ \) would require the quadriceps to generate a knee extending moment of approximately 10 Nm per leg (\( M_b = 50 \text{ kg}, \ l = 0.50 \text{ m} \)). In the case of increased \( \theta \) (e.g. due to mechanical obstructions in the knee joint) or increased \( T_k, \text{pas} \), this moment may easily be higher. If the quadriceps condition is insufficient to counteract the required knee moment, \( T_k \) can be reduced by increase of \( T_a \). This may be done by bracing of the ankle joint or by stimulation of ankle plantar flexors. \( T_a \) is limited to approximately \( T_a \leq \frac{1}{2} M_b g L \) per leg. This indicates a fixed ankle joint to stabilize the knee joint without the need of stimulating knee extensors for values of \( \theta \) which are given by:

\[
\frac{1}{2} M_b g l \sin (\theta) + T_k, \text{pas} \leq \frac{1}{2} M_b g L
\]

When \( T_k, \text{pas} \) can be neglected (no contractures or mechanical knee bracing) this gives a stable knee joint for approximately:

\[
\theta \leq \arcsin \left( \frac{L}{l} \right), \quad \text{with} \ 0 \leq \theta < \pi/2.
\]

METHODS

Subjects

The three patients in this study had complete, traumatic lesions from T5 to T7, with no evidence of significant peripheral nerve damage. All had participated in an FES muscle training programme as described by Mulder et al. and similar to that used by Kralj et al. for at least 18 months at the time of the experiments. The patients were aged between 20 and 29 years and did not show any significant contracture or spasticity and had been using LSU type reciprocating gait orthoses for walking on a regular basis at home.

Protocol

Finite state control was compared with open-loop control of knee extensor muscles, both with a free ankle and with the ankle joint mechanically stabilized. These two conditions were derived from the daily life and clinical situation in FES standing. For each subject a series of four standing experiments was carried out, standing being performed within a frame allowing the patient to use his non-paralyzed upper body musculature for maintaining balance. The frame was strain gauge instrumented to measure the arm forces during standing, and experiments were separated by 1.5 h of rest, during which the patients were in their wheelchairs; they were asked to be moderately physically active in order to facilitate quadriceps recovery. Stimulation electrodes were not removed between experiments. The following standing experiments were performed:
1. open-loop knee control, ankle joint free;
2. artificial reflex control of knee joint, ankle joint free;
3. open-loop knee control, ankle mechanically stabilized;
4. artificial reflex control of knee joint, ankle mechanically stabilized.

Preceding the four regular experiments one dummy experiment was performed to obtain equal initial muscle conditions throughout the protocol. For the same reason stimulation was continued after experiments 2 and 3 (usually in the sitting position) until muscle force equaled the force on termination of experiment 1.

Each standing experiment was preceded and followed by measurement of maximum knee extensor torque as derived from the static stimulus amplitude to knee torque relation (recruitment curve). This was performed bilaterally while the patient was sitting. Each standing session thus consisted of: (1) recruitment measurement during sit, (2) standing in the instrumented standing frame, and (3) second recruitment measurement during sit. The three measurements were separated by a maximum of 15 s. Standing was continued until fatigue resulted in knee buckling exceeding 20° (at which the duration of standing was assessed), unless standing duration was more than 10 min. This time was selected for practical experimental reasons.

During standing, hip and trunk were stabilized using the above-knee part of the patient’s regular reciprocating orthosis. Additional ankle bracing in experiments 3 and 4 was obtained by adding the below-knee part of this brace. In all experiments the knee joints of the brace were removed to allow the knee joint of the patient to move freely without friction.

The controller

Two independent controllers were used for each knee joint separately. Each controller had the finite state rule-based structure as shown in Figure 2, and as in a preliminary form previously documented by Mulder et al.1. During lock, stimulus amplitude was decreased exponentially (time constant typically 1 s). During unlock (which is signalled when knee angle deviates from the locked position more than a predefined threshold (typically 1.8°)) stimulus amplitude was switched to supra maximal to return the joint to the locked position. To detect this transition to the locked state angular velocity was used, allowing the system to calibrate automatically to the locked position after each excursion to the unlocked state; the zero velocity situation was independent of any DC shift in goniometer signal. Calibration was also performed when the knee was further extended while being in the locked state, which may indicate goniometer disturbance.

The controller was implemented for both knee joints individually using a Tulip-AT computer with AD facilities (Analog Devices, RTI-815, 12 bit) and a multi-channel digitally controlled high-output impedance current stimulus developer at our laboratory (monophasic, rectangular pulses). The quadriceps were stimulated using an adhesive surface cathode and anode (Pals, Axcelgaard Manufacturing Co. Ltd., Fallbrook, CA, USA, 4 x 8 cm) placed over the motor points of rectus femoris/vastus lateralis and vastus medialis respectively. Pulse duration was fixed at 300 μs, pulse rate was set to the minimum frequency for a fused contraction: 20 Hz. Knee angle was measured using an externally mounted goniometer (MCB pp27c, 310°, nonlinearity 1%). To determine knee angle (bandwidth 10 Hz), the goni-signal was sampled at 100 Hz and digitally first-order low-pass filtered with a cut-off frequency of 15 Hz. Angular velocity was calculated from the knee angle inter-sample difference and smoothed by a digital third-order Butterworth low-pass filter with a cut-off frequency of 15 Hz.

**Experimental protocol**

Standing was performed in a strain gauge instrumented frame, consisting of two short single end-mounted parallel bars which were adjustable in height. Before and after standing, isometric recruitment data of the quadriceps muscle were recorded. This was done while the patient was sitting, simultaneously from both legs using a bridge configuration of strain gauges mounted on two rigid steel bars connected to bridge amplifiers (Phillips, PR 9307). Recruitment was measured by applying ramp-up and ramp-down of stimulus amplitude (0 to 100 mA; 10 s up, 10 s down). All data were stored on disk for off-line evaluation. Sampling rate was 100 Hz in all cases.

**RESULTS**

A typical response showing the performance of the finite state controller during standing is shown in Figure 3. The patient was standing with the ankle joints freely moving (type 2 experiment). Figure 3a shows stimulus amplitude, knee angle and vertical armforce in response to a large anterior/posterior sway of the body (typically 20°) voluntarily induced. At the onset of the disturbance the stimulator responded immediately by switching on the stimulation. The changed posture then resulted in a hyper-extension of the knee joint and increase of arm force and stimulation decreases. At the following angular
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Figure 3 Finite state control of the knee joint during standing. Subject JM, experiment type 2, right leg, ankle joint freely moving. Shown are two 8-second traces of stimulus amplitude (1), knee angle (2) and vertical arm force (3) from one experiment. 180° corresponds to full knee extension. a, Responses to an anterior/posterior body sway of approximately 20° induced voluntarily by the patient at t = 93 s; b, responses to natural disturbances during stable standing. Stimulation is switched on when knee flexion exceeds a certain threshold relative to the (local) maximum knee angle value which (locally) is considered as the locked position (indicated by arrows)

flexion (body back to neutral position) stimulation is switched on repeatedly to stop this movement until a new stable position is reached. During stable standing (Figure 3b) artificial reflex actions (switch-on of stimulation) were also elicited regularly. On average only a small amount of knee extensor activation was required in the specific experiment. As the ankle angle $\theta$ was in this experiment approximately zero during stable standing, either $T_{k,\text{max}}$ exceeds $T_0$ in the particular patient (equation 1), or external disturbances bring the knee out of the stable position frequently. In general, large variations were seen over the time of an experiment with periods of frequent switching of stimulation and periods of less stimulation.

Figure 4 shows knee flexion and the maximal locking velocity throughout the standing experiment of Figure 3. The first flexion angle exceeding 20° was taken to determine the duration of standing (324 s for the specific experiment). Although the patient could reestablish the knee by using her upper body, standing was unstable from that moment, requiring increased arm effort and showing larger perturbations of the knee angle.

The following parameters were determined to quantify the standing performance of the patient under the different experimental conditions (Table 1): duration of standing ($D$), relative loss of maximum torque at the knee as determined from recruitment data ($TL$), average of maximal knee flexion angles as recorded during each unlock ($FL$), average of maximal locking velocity during each unlock ($LV$), average stimulus amplitude ($ST$) and average vertical arm force ($FA$). All averaging was over the period that stable standing was obtained, and did not include standing up. As a reference for the average stimulus amplitude the threshold ($THR$) of the quadriceps was also determined. When the average stimulus amplitude is below the threshold this indicates the muscle to be off for long periods.

For all subjects finite state control (experiments 2 and 4) resulted in longer duration of standing or (when duration of standing was equal to the maximal

<table>
<thead>
<tr>
<th>Subject</th>
<th>Exp</th>
<th>$D$ (s)</th>
<th>$TL$ (%)</th>
<th>$FL$ (°) ± SD</th>
<th>$LV$ (°/s) ± SD</th>
<th>$ST$ (mA)</th>
<th>$THR$ (mA)</th>
<th>$FA$ (%) ± SD</th>
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<tr>
<td>JM</td>
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<td></td>
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<tr>
<td></td>
<td>2</td>
<td>324</td>
<td></td>
<td>5.2 (8.9)</td>
<td>10.7 (12.2)</td>
<td>27.3</td>
<td>30</td>
<td>12.2 (13.3)</td>
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<td>3</td>
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<td>84.2</td>
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<td>600</td>
<td>15.9</td>
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<td>17.9</td>
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<td>29.4 (14.0)</td>
<td>20.7</td>
<td>31</td>
<td>3.4 (4.0)</td>
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</tbody>
</table>

$D$: stand duration (time until knee buckling exceeds 20°, maximum 10 min)

$TL$: relative torque loss ([maximum torque before standing]−[maximum torque after termination of standing])/[maximum torque before standing]

$FL$: average maximum knee flexion during unlock

$LV$: average maximum angular velocity during knee locking

$ST$: average stimulus current amplitude

$THR$: threshold of the muscle

$FA$: average vertical arm force (in percent body weight)
time of 10 min) in less torque loss in the quadriceps than when applying open-loop control (experiments 1 and 3). Differences of a factor 2.5 to 12 in stance duration and of a factor 1.5 to 5 in torque loss were found. Average stimulus current amplitude \( ST \) in case of finite state control was lower than the threshold of the muscle for all experiments except DV, 2. Although the threshold was determined under different conditions (non-extended situation, knee flexion 30°), lower stimulus amplitude suggests that the muscle was off for major periods of time, indicating a net flexing knee moment near zero in most experiments. Knee flexion angle during unlock \( FL \) on average was 2.0 times lower with ankle bracing (experiment 4) than without (experiment 2) indicating increased postural stability when using ankle bracing. Arm support showed a similar relation to ankle bracing and on average decreased from 13.9% (experiments 1 and 2) to 7.5% body weight (experiments 3 and 4). On the other hand, ankle bracing had a negative effect on average locking velocity in two of the three patients.

DISCUSSION

Several strategies have been proposed in order to minimize the stimulation required for the quadriceps in FES controlled paraplegic standing\(^1\).\(^3\).\(^4\).\(^5\). Some of these strategies were based on the detection of the knee joint being stabilized by external forces, switching stimulation off when the ground reaction force passes in front of the knee joint axis\(^1\).\(^7\). In other studies quadriceps activation was reduced based on e.g. feedback of knee angle information\(^4\).\(^3\). From the latter studies calibration of the angle measuring system and inducing an abrupt increase of quadriceps activation in response to angular displacements have been reported to be important. However, the potential of such a system was demonstrated by Marsolais et al.\(^5\) reporting 35 to 47% decrease in total energy consumption in paraplegic standing when muscle activation was reduced to the minimum value required for standing. The method of force reduction proposed by Mulder et al.\(^2\), and evaluated clinically in the current study, basically uses the second approach and reduces knee extensor activation without the need for external stabilization devices. In addition it fulfills the demands for repeated calibration of the angle sensory system and of an abrupt response to angle perturbations.

For all subjects, finite state control resulted in appreciably higher duration of standing or appreciably less torque loss in the quadriceps than when applying open-loop control. When duration of open-loop standing is corrected for the differences in initial knee extensor torque (by estimating the open-loop standing duration from the individually known quadriceps responses during sustained contraction\(^1\) the ratio of the standing durations in open- and closed-loop experiments is 6 for subject JM and 5 for subject DV.

Finite state control was found to work satisfactorily both with and without the use of external ankle stabilization. Our results indicate that in some patients no ankle bracing is required to obtain long-duration standing. This is indicated by the results of subject RD, experiment 2. In other patients (like JM and DV) the improvement of standing duration which results from finite state control may be insufficient to be functional. In that case ankle bracing may be used to stabilize the knee joint from ground reaction force without basically affecting the control strategy. Also in some patients ankle bracing may be required for the sole purpose of protecting the ankle joint in the lateral direction.\(^1\) However, a fixed ankle joint may affect the performance of an FES system during ambulation as it limits the possibilities for active (FES-induced) push-off and may cause increased knee loading during heel strike. Therefore, should ankle stabilization be required (e.g. in case of insufficient knee extensor condition), the bracing should preferably stop dorsal flexion, meanwhile allowing plantar flexion.

As could be expected from equation 2, external ankle bracing did not always result in average stimulus amplitude being zero during finite state control. When the ankle joint is stabilized, small posterior displacements of GRF, as caused by posterior movements of the trunk relative to the neutral position, cause the knee joint to flex proportionally to these trunk movements. As the lower leg would also be stable without stimulation under these circumstances, sensitivity of the system for these knee angle disturbances may be decreased e.g. by increase of the angular threshold of the controller.

One of the characteristics of the finite state (or artificial reflex) controller is the presence of a limit cycle oscillation at the knee joint, which may have bearings on both the convenience and the confidence of the patient whilst standing. Maximum average
knee flexion during unlock was 7.2° (RD, experiment 2). For an average person this corresponds to a vertical hip movement of 2 mm which was not found to cause major inconvenience. However, it is the subjective impression of stability which finally governs the acceptance of the control strategy for the patient. From that view it is interesting that in all experiments average vertical armforce was lower with ankle bracing than without, as was the standard deviation. This correlates to the improved stability of the knee joint under ankle bracing conditions, although difference in stability was stated by only one of the three patients when asked.

The question of whether the controller actions will influence the condition of the ligaments of the knee joint is difficult to answer. For traditional open-loop stimulation the issue of possible damage from stimulation is still under investigation. Although locking velocity was relatively low during finite state control, care must be taken. When needed, an extension-stop type knee brace can be applied easily without affecting the control strategy.

During standing the hip joints were stabilized mechanically. If the hip were not stabilized the resulting C-posture (hip hyperextended up to typically 45° to lock the hip on its ligaments, resulting ankle angle typically 15°) would require a minimum quadriceps moment of 35 Nm per leg (equation 1). This is 70% of the average 50 Nm which can be produced after thorough FES muscle conditioning and this should be avoided as it may be expected a priori to limit muscle endurance. In our study mechanical stabilization of the hip was preferred over stimulation as it is a convenient way to immobilize the trunk as well.

Considering the application of FES-induced walking, the proposed control strategy is expected to be useful during the stance phase. In that case it may be advantageous for the controller intrinsically to allow for adjustable initial knee flexion at heelstrike before knee extensor activation is switched on. This may lead to a more natural gait in comparison with traditional open-loop applications where the quadriceps are fully activated before heelstrike.

REFERENCES


