An Apparatus for FES-assisted Arm-Cranking Exercise in Tetraplegia

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Abstract

We describe a novel system for FES-assisted exercise for upper-arm muscles in C4-C6 tetraplegia. A commercial arm-crank ergometer is interfaced with a neuromuscular stimulator to provide FES to the biceps and triceps muscles for arm ergometry. The design is described and initial results are shown which illustrate the setup and demonstrate the feasibility of our approach as a means to exercise upper-arm muscles in tetraplegia.

1 Introduction

Spinal cord injury in the cervical (neck) region can result in paralysis of the upper and lower limbs. A person with an injury at level C5 or C6 will generally retain control of the shoulder and the elbow flexor muscles (biceps), but will have no control of the hand or of the elbow extensor muscles (triceps). In complete C4 injury control of the entire arm is lost.

The use of Functional Electrical Stimulation (FES) in C4–C6 tetraplegia has focused principally on open-loop systems for hand function [4] and improved working area (i.e. overhead reach) [1].

The potential for the provision of upper-limb exercise modalities using FES assistance has been identified by Needham-Shropshire et.al [3]. They used FES of the triceps muscles in C5/C6 individuals to perform arm-crank ergometer (ACE) exercise. A passive ergometer was used in their study. The evaluation of the triceps muscle grade showed that a significant improvement could be achieved through FES-assisted exercise.

We have developed a system which allows people with C4–C6 tetraplegia to do arm cranking exercise assisted by FES of the triceps *and* biceps muscles. An active ergometer is used which allows both active and passive exercise regimes. While C5/C6 patients *are* able to do arm cranking without FES, voluntary muscle weakness and the absence of elbow extension torque means that this is highly ineffectual. The addition of triceps via FES, and the stimulation of biceps which are only partly under voluntary control can, we propose, greatly improve the quality of cyclic upper-limb motion. We focus on quantitative outcome measures such as crank moment and generated power output, and we will evaluate the cardio-pulmonary response to exercise.

This paper focuses on the technical description of our system and its implementation. Initial results from experiments with one subject are used to illustrate the approach and to show its feasibility.

2 Methods

The overall setup is shown in Figure 1 and consists of the ACE device, the pattern generator and the neuromuscular stimulator.

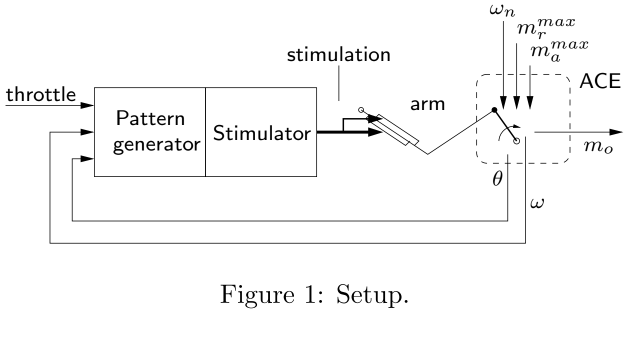

Figure 1: Setup.

The ACE device (MOTOmed viva, Reck medical devices, Germany) provides measurements of the crank angle θ, the angular velocity ω and the motor torque m_o (see below) via a serial communication interface. The crank angle and the angular velocity are used in the pattern generator to decide when each muscle group is to be stimulated. The stimulation intensity is set by a "throttle" which is implemented as a potentiometer. The pattern generator drives the neuromuscular stimulator (Stanmore Stimulator) via an isolated RS232 link. Four stimulation channels are used: left and right biceps and left and right triceps. Stimulation is applied using standard surface electrodes (Pals, Axelgaard), the stimulation frequency is 20Hz. The extension and flexion moment induced by the stimulation at the elbow generates a force at the cranks of the ACE device which induces an external moment, m_e.

The ACE device has three input parameters which can be set through its display terminal: the nominal crank velocity ω_n, the maximal resistive moment m_r^{max} and the maximal active moment m_a^{max}. Without external moment at the cranks, the ACE device will drive the cranks at the nominal angular velocity, ω_n. If the external moment m_e acts as a load, then the nominal speed will be maintained as long as $|m_e|$ does not exceed the maximal active moment, m_a^{max}. If $|m_e| > m_a^{max}$, the velocity is going to decrease. If the external moment acts in the same direction as the crank movement, the device will resist acceleration above ω_n with a maximal resistive moment m_r^{max}. If $|m_e| > m_r^{max}$, the velocity of the cranks increases.

Figure 2: Subject exercising.

The measured motor torque m_o is calibrated in such a way that it is zero when no external force is applied. If an external resistance is applied at the cranks and the motor has to generate an active torque, then $m_o > 0$. If the cranks are propelled actively by the subject and the motor has to generate a resistive torque then $m_o < 0$.

Figure 2 shows a subject exercising in the ACE device.

The pattern generator uses an algorithm similar to that developed for FES-assisted cycling [2]: For each muscle group, a nominal angular stimulation range is defined. These are shown in Figure 3. To compensate for the delay between stimulation and muscle contraction the stimulation ranges are brought forward as crank speed increases.

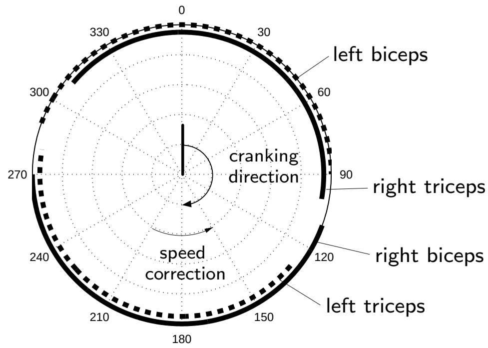

Figure 3: Nominal stimulation ranges. The angle θ is shown as the position of the right crank arm.

Pilot experiments were conducted with one tetraplegic subject. The subject is 17 years old, 6 months post injury, level C4 incomplete. All four muscle groups show strong reaction to surface stimulation. The voluntary contraction of the triceps and biceps muscles is of grade 3 or lower (Oxford scale) with no proprioception in the upper limbs.

3 Experimental Results

In our initial experiments the active moment was set to $m_a^{max} = 5$Nm, the resistive moment is $m_r^{max} = 1$Nm and the nominal velocity is $\omega_n = 5$rpm. The stimulation current was set to 30mA for all channels, and the pulsewidth was varied between 0 and 200μs. The total output moment was measured for a number of cycles and then averaged for the range of one cycle ($0 - 360^o$). Results of measurements obtained with maximal stimulation and without stimulation are shown in Figures 4 and 5. The angle θ is shown as the position of the right crank.

The results without stimulation show a large motor moment m_o between 330^o and 90^o and (symmetrically) between 150^o and 270^o. This means that in these ranges a significant active moment from the ACE device is required to move the arms. These ranges correspond to the situations when both arms are flexed and thus both elbows simultaneously rub slightly

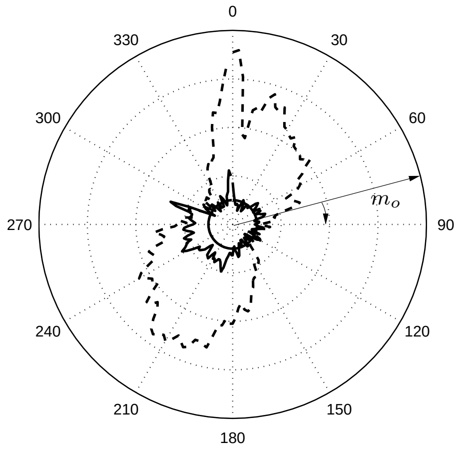

Figure 4: Motor moment m_o as a function of the crank angle; polar plot. The radial distance of any point from the centre indicates the instantaneous magnitude of m_o. The solid line shows results with stimulation (pulsewidths $200\mu s$), the dashed line shows measurements with no stimulation. The solid circle in the middle denotes $m_o = 0$.

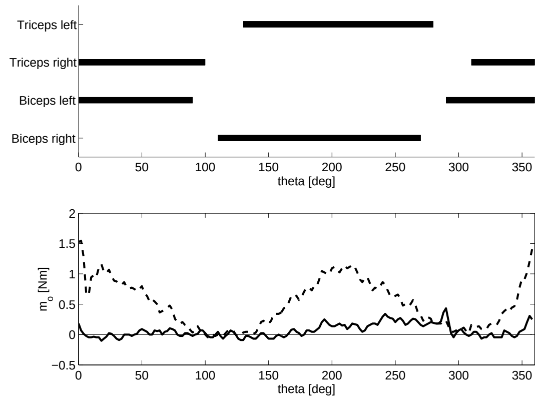

Figure 5: Top graph: stimulation patterns vs. crank angle for the four muscle groups. Bottom graph: motor moment as a function of the crank angle. The solid line shows results with stimulation (pulsewidths $200\mu s$), the dashed line shows measurements with no stimulation.

against the upper body. In the ranges where the motor moment is minimal (around 120^o and around 300^o), one arm is fully extended providing minimal resistance for cranking.

The results with stimulation show a clear decrease in the motor moment m_o. The moment difference to cranking without stimulation is provided by active muscle contraction through FES. The motor moment is, however, only occasionally smaller than zero, indicating that arm cranking with FES alone would not currently be possible for this subject without active support by the ergometer.

4 Summary and Conclusions

An active arm-crank ergometer was interfaced with an FES system for stimulation of the biceps and triceps muscles to allow for FES-assisted ergometry. We have shown that the approach is feasible: a subject with very limited voluntary control of the biceps and triceps muscles is able to produce significant moments with stimulation. The generated moment varies significantly with the position of the cranks, and it only suffices for a part of the range to propel the cranks. Using an active ACE device, however, allows performance of the exercise even in this case as the cranks are moved actively by the machine through the difficult parts of the cycle. The methodology is therefore suitable as an effective means to exercise upper-arm muscles in tetraplegic individuals.

Based on this approach, we are now proceeding to a full pilot study which is aimed at evaluating the effectiveness of this exercise modality based on quantitative outcome measures (such as crank moment and output power) and measurements of the cardiopulmonary responses.

Acknowledgments

This work is supported by the UK Eng. and Phys. Sci. Research Council (Grant GR/R30730) and by the European Commission (Grant MCFI 2000-02190).

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