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Doctoral School of Biology and Sportbiology

Cyclic limb movements by natural and artificial control

The effect of external load on arm cycling variances of able-bodied participants and FES assisted lower limb cycling of spinal cord injured patients

PhD thesis

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PÉCS, 2022

1. INTRODUCTION

In my dissertation I write about the control of cyclic limb movements: the arm cycling in able-bodied participants and the functional electrical stimulation (FES) assisted lower limb cycling in spinal cord injured participants.

Arm cycling on arm ergometers is often applied in sports training when the aim is to strengthen upper body muscles in neurologically intact individuals (Elmer, Danvind, et al., 2013) or to assess muscle powers and evaluate performances (Hübner-Wozniak et al., 2004). Arm cycling exercises are also included in medical rehabilitation protocols (Zhou et al., 2018) to improve motor performance and motor control of individuals with spinal cord injury (SCI) or stroke (Lasko-McCarthey & Davis, 1991; Zehr et al., 2012). Despite a range of sport and rehabilitation applications (Elmer, Marshall, et al., 2013; Matjačić et al., 2014), the literature on arm cycling movements is limited relative to that on lower limb cycling. However, the importance of arm cycling has recently been supported by several investigations (Botzheim et al., 2021; Elmer, Marshall, et al., 2013; Laczko et al., 2016; Matjačić et al., 2014; Mravcsik et al., 2016). It has also been shown that arm cycling training improves strength, coordination of muscle activity during other types of motor tasks, such as walking, and neurological connectivity between the arms and the legs (Kaupp et al., 2018). In arm cycling, we examined the dependence of movement patterns on body position and movement size and found that body position has a significant effect on muscle coordination (muscle synergies), but the size of the movement has not (Botzheim et al., 2021).

My particular purpose was to analyze physiological parameters of human subjects during arm cranking when an altered load is applied (effect of crank resistance). The metrics we analyzed are the arm configuration variance (in 3D joint space) and the muscle activation variance (in 4D muscle activation space). These metrics can indirectly validate the type of control utilized for this complex task.

The arm configuration is given a “joint space”, the dimension of this space is equal to the considered number of joints. Thus, the arm configuration is given by a vector whose coordinates are the intersegmental joint angles in the shoulder, elbow and wrist. The variance of such arm configurations were investigated in reaching arm movements. (Domkin et al., 2002; Kang et al., 2005) and in those movements in which an object held in the hand, had to be transposed form one place to another one. The variance of arm configurations depended on the weight of the object held in the hand only when the object was transposed vertically upward from a lower to a higher position. Beside the investigation of arm configurations, the variance of muscle activations has also been examined. It was found that the variance of muscle activation always depended on weight of the object in the hand. (Tibold et al., 2011).

Arm muscle activities during arm cycling at different workloads were characterized in (Chaytor et al., 2020) and it was found that there was a linear relationship between EMG amplitude and power output for individual muscles. However, it has not been studied how the cooperation of many muscles and the variance of these muscle activation depend on the resistance of the crank. I examine how the variance of muscle activation changes as the resistance of the crank changes. Why is this important? The effect of crank resistance on arm configuration variance and muscle activation variance is an interesting and important issue from a movement control perspective. If the joint configuration variance is not affected by crank resistance, it suggests the separation of kinematic- and force-control (Kolesnikov et al., 2011; Piovesan et al., 2019) where the kinematic task can be maintained safely when crank resistance is altered. Knowing the type of control strategy is important in training and rehabilitation protocols.

Arm cycling exercises and the application of functional electrical stimulation (FES) induced lower limb contractions in the same time have a positive effect on cardiovascular function in SCI patients (Davis et al., 1990). Arm cycling exercises are also used in combination with functional electrical stimulation (FES) training of individuals with spinal cord injuries. (Brurok et al., 2013; Bakkum et al., 2015). In my dissertation I write about FES-assisted lower limb cycling.

Functional electrical stimulation is a rehabilitation procedure in which we create a functional movement, with electrical impulses. People with paralyzed lower extremities, this is the only way to activate the paralyzed muscles – to generate muscle contraction based on predefined pattern. Our research group is the only one in Hungary to perform measurements at the National Institute of Medical Rehabilitation on the topic of FES-assisted lower limb cycling.

There are several reasons for spinal cord injury. The active age population is most affected (Ahuja et al., 2017; Devivo, 2012). Most of it is traumatic reason - road accident, falling from a high place, violence. The proportion of male injuries is higher (Chen et al., 2016; World Health Organization, 2013). The number of new injuries in Hungary is about 350-550. Spinal cord injury can result in loss of motor, vegetative and sensory functions. The positive effect of FES have been reported in a number of studies. The positive effect on quality of life was examined with questionnaire (Dolbow et al., 2013). SCI leads to muscle atrophy, osteoporosis and stiffening of the bones. The amount of load on the bones and the effects on the forces are reduced. Femur and tibia are most commonly affected by osteoporosis, with bone density decreasing by 50% but even 70 % (Eser et al., 2004; Frotzler et al., 2009). Trabecular bone density and total bone density of the distal femur epiphysis were significantly increased by FES, but no significant difference was measured in the proximal and distal epiphyses of the tibia (Frotzler et al., 2008). The limited aerob capacity of SCI individuals contributes to an increased cardiovascular risk profile (Bauman & Spungen, 2008). This limited aerob capacity is related to the height and extent of the injury. FES cycling is an effective way to improve decreased heart capacity and increase maximal oxygen uptake (Brurok et al., 2013).

In some cases of spinal cord injury, we speak about denervated muscles – the SCI is below T10 or in the cauda equina region, supplemented by a lower motoneuron lesion. The excitability of the denervated muscles, the sensitivity of the electrical membrane, strongly depends on the degree of degeneration or regeneration of the muscle, but it is certainly much lower than the sensitivity of the neuron. Stimulation of such muscle fibers requires a pulse duration of 10 to 150ms, a longer period of time after severe degeneration to induce depolarization on the muscle membrane at a given location along the muscle fiber, followed by action potentials that may travel either direction toward the ends of the muscle fiber (Mayr et al., 2002).

2. SPECIFIC AIMS

2.1. Arm Cycling

When cycling with the arm, the hand moves on a specific, fixed circular path, while the arm can take different poses, forming different arm configurations in each lap. Theoretically, an infinite number of changes in the intersegmental angles between body parts at the shoulder, elbow and wrist can result in hand movement in a given path. A wide variety of coordination of joint rotations is possible. One of the aims of my research is to explore how, in practice, this diversity depends on an external force, namely the resistance of the crank of an ergometer. We asked the following questions:

How does the variance of the arm configurations change during the arm cycling movement if the resistance of the ergometer crank changes? So if the hand is constrained to a fixed path but the load on it alters.

In addition to various arm configurations, an infinite number of combinations of muscle activities and their changes can result the movement of the hand in a given path. Another goal of my research was to answer the following question:

How does the variance of muscle activations during arm cycling depend on the resistance of the arm-cycle ergometer?

2.2 FES-controlled lower limb cycling

Another goal of my work was to design training protocols and conduct FES induced cycling training for spinal cord injured patients at the National Institute of Medical Rehabilitation. My goal was to examine the effect of the training on the physical performance and energy values, physical characteristics, and heart rate and blood pressure values by answering the following questions:

Are the effects of the same muscle stimulation pattern on physical performance different in different patients?

Do two different stimulation patterns have the same effect on the physical performance of the same spinal cord injured person?

How do heart rate and blood pressure change during training?

What improvement can be achieved in the mean distance traveled and speed values of FES-induced cycling movements in patients with denervated muscles?

3. METHODS

3.1 Arm Cycling

The arm cycling movements of fifteen able-bodied persons were analyzed by kinematic and muscle activity measurement. Their mean age was 24 ± 4 years, including 7 men and they were all right-handed. The participant was seated in a fixed chair in front of an arm-cycle ergometer (MEYRA, Kalletal, Germany). The movement task was arm cycling against 3 different crank resistances (3 resistance conditions). For recording kinematic and muscle activity data a Zebris CMS-HS ultrasonic motion analyzer system was applied, with sampling frequency of 100 Hz for kinematic data and 900 Hz for electromyographic (EMG) data. Simultaneously, 8 muscles' electrical activities (M. Biceps Brachii (BI), M. Triceps Brachii (TR), M. Deltoideus anterior (DA) and Posterior (DP) muscles in both arms) and positions of 8 kinematic markers were measured (M1: Reference marker on the back of the chair, M2: Acromion of the shoulder, M3: Distal end of the upper arm (epycondilus lateralis humeri), M4: Proximal end of the forearm (caput radii), M5: Styloid process of the ulna) , M6: Radius styloid process of the radius, M7: Little finger (caput of the fifth metacarpal bone), M8: Reference marker on the crankshaft). At least 30 laps had to be completed under each cycling condition, and the speed was given as 60 rpm (revolutions per minute), supplemented by a metronome.

The recorded data were processed with Matlab and Excel programs. To filter the EMG data, we used a type 3 Butterworth filter with a lower frequency limit of 25 Hz and an upper frequency of 300 Hz, as well as a 50 Hz component corresponding to the mains frequency, and smoothed with RMS (Root Mean Square), with 80 sampling wide window. Time series of marker coordinates were filtered by discrete cosine transformation (DCT) (Shin et al., 2010). Data processing was continued by calculating intersegmental angles, which were calculated from the coordinates of the markers for all three joints: shoulder (α), elbow (β) and wrist (γ).

Time series containing muscle activities and joint angles were divided into cycles (circles) to compare the muscle activities and joint angles which where measured in each cycle.

Finally, we calculated the arm configuration variance:

$$V_{ang}(t) = \frac{\sum_{k=1}^{N}[\bar{a}(t) - a_k(t)]^2}{N * 3}$$

where $a(t) = [\alpha(t), \beta(t), \gamma(t)]$, and $t = 1, \ldots, 100$ (normalized time: the cycling time for a complete circle is divided into one hundred equal parts). Overline indicates average. k denotes the serial number of the circle, N denotes the number of circles and a multiplier of 3 denotes the degree of freedom (dimension of the vector α(t)) since three joint angles are taken into account.

Muscle activation variance:

$$V_{EMG}(t) = \frac{\sum_{k=1}^{N}[\overline{Mus}(t) - Mus_k(t)]^2}{N * 4}$$

where $Mus(t) = [BI(t), TR(t), DA(t), DP(t)]$ and $t = 1, \ldots, 100$ (normalized time: the cycling time for a complete circle divided into one hundred equal parts). A multiplier of 4 denotes the degree of freedom (dimension of the vector Mus (t)) because the activity of four muscles is taken into account. Thus, the combined activity of the four muscles is characterized by the Mus (t) vector.

These two variances were averaged separately, so the time averaged for each normalized time point was also calculated for arm configuration and muscle activation. Thus, the variance of arm configuration and muscle activation in each cycling condition can be characterized by a number:

$$\bar{V_x} = \frac{\sum_{t=1}^{100}\left(V_x(t)\right)}{100}$$

where $x = ang$ or $x=EMG$.

Multiple ways mixed factor analysis of variance (ANOVA) was used for statistical analysis to interpret EMG and kinematic data. Tukey's post-hoc test was used to examine the effect.

3.2 FES controlled lower limb cycling

The participants were selected from the Rehabilitation Department of Spinal Cord Injuries of the National Institute of Medical Rehabilitation, with a medical permit or recommendation. The ethical permit was given by the decision of the Ethics Committee of the National Institute of Medical Rehabilitation dated 4 October 2017. I present the results of four different comparisons in the results section, in which the participants differed. Spinal cord injured patients performed FES controlled leg cycling on a Motomed Viva 2 bicycle ergometer that was connected to an eight-channel muscle stimulator developed at Pázmány Péter Catholic University. Stimulation was performed with a pulse width of 300μs and a frequency of 30Hz, the current form used was monophasic, trapezoidal. Stimulation patterns (as a function of the ergometer's crank angle) were determined based on healthy movement patterns. The quadriceps and hamstring muscle groups were stimulated with a pattern named H2, or the vastus lateralis, vastus medialis, rectus muscles and biceps femoris were stimulated a pattern that was named H4.

Participant sat in a wheelchair, the ergometer was placed at the front of the wheelchair and the participant's feet were placed on its pedals. Active cycling lasted up to 30 minutes. Blood pressure was measured at the beginning of the training, after cool down and every 8-10 minutes during the training. Mean arterial pressure was calculated, as well as the mechanical power and energy delivered for each training session.

Spinal cord injuries with denervated muscles drove a tricycle with their paralyzed muscles. The tricycle was equipped with a 4-channel electric muscle stimulator, developed at the Medical University of Vienna, which was able to emit the long-term pulses (15-100 ms, +/- 80V) needed to activate the denervated muscles. Two channels were used, with large (200 cm^2) surface electrodes (Schuhfried Inc, Vienna Austria) placed on the quadriceps muscles in a wetted pad to stimulate and activate them for knee extension. The seat of the tricycle is placed on a rail, sliding back and forth during movement. By stimulating the quadriceps, the knee extends, the seat slides backwards and while driving the rear wheels. Importantly, this is not a static tricycle driven on wheels in one place, but capable of moving.

4. RESULTS IN ARM CYCLING

We found that in arm cycling, the variance of the arm configuration is not affected by the resistance of the crank at either the dominant or non-dominant arm, nor when cycling with one arm or two arms, whereas when the resistance changes, muscle activation variance changes quadratically with respect to the change in muscle activation.

The shape of the variance time profile does not change after the change in resistance, either in terms of variance in arm configuration or variance in muscle activation. There are three maxima of variance, at approximately 90º, 180º and 270º crank angles when zero means that the crank of the ergometer points toward the cyclist. At 180º, the variance reaches its absolute maximum, where the elbow is most stretched. In the statistical analysis, we found no significant difference in either the factors or the variance in the arm configuration. At the same time, we found a difference in the degree of crank resistance and the mode of rotation in the muscle activation variance. Furthermore, there is a significant difference / interaction between the subject and the side (left/right arm) and the subject and mode, indicating that the subjects are statistically significantly different from the two sides during the task, and we also found a significant difference between the two-handed and one-handed cycling. All this suggests that the subject should be considered an independent factor.

The magnitude of the changes in the intersegmental joint angles only slightly differed from each other at different crank resistances. This was observed for the shoulders, elbows and wrists as well.

5. RESULTS IN FES DRIVEN LOWER LIMB CYCLING

I divide my results in the field of FES-induced cycling into four parts. I describe the effects of a given stimulation pattern on several spinal cord injured participants, then the effect of two different stimulation pattern on one participant, and hen the results obtained during the training of ten patients. Finally, I write about the results of tricycling movements by denervated muscles.

The same stimulation pattern (H2) was used in 3 patients whose level of injury, the cause and severity of their injury was different. My studies show that performance values (power output and energy output) increased during the training in the case of each participant, even if at different extents.

14–14 training sessions of two participants were controlled with the same stimulation pattern (H4), where we found that the first subject had significantly higher mechanical power output and energy output. We concluded that the difference was caused by the level of the injury (the injury level of the first participant was Th8 (polytraumatic), that of the second participant was C5-6).

In a spinal cord injured patient, both stimulation patterns were used for 10-10 training sessions. Our results show that the subject achieved significantly higher performance with the H4 pattern.

I examined the average heart rate and mean arterial pressure values measured during the training of 172 FES lower limb cyclists. Each subject had a minimum of 10 trainings. We found large differences in the individual values. The change in heart rate after the start of the training varies from individual to individual, but in general it can be said that after the end of the training a decrease was measured, at the cool-down phase of the training the heart rate was at the resting level. The mean arterial pressure rises at the beginning of the training and then remains at this elevated level. In the cool-down phase of the training, its value decreases, from which we can deduce a good training effect.

Two denervated spinal cord injured patients used the stimulator-equipped tricycle twice a week for six weeks. The distance traveled during the 12 training sessions increased from an initial 2.23 km to 3.33 km for the first and from 1.61 km to 2.25 km for the second participant. The average speed also showed an increase. Within a training session, the speed drops in the middle of the session (and thus the distance traveled) but in the last phase, the participants increase the speed again. In spinal cord injuries with denervated muscle, FES has not yet been used to generate locomotion of the patient. My work is the first which shows that patients with this type of injury can achieve results.

6. SUMMARY

In my dissertation, I showed that the variance of the configuration of the human arm does not change with increasing the resistance of the ergometer crank during arm-cycling movement of able-bodied participants. In principle the task could be performed with an infinite number of arm poses and the hand could be moved in a given path, but this variance does not change as the external resistance changes. Our results show that in the case of arm cycling movement, if we increase the crank resistance, the variance of the magnitude of the muscle activation vector increases, but its time profile does not change. From the above, I concluded that central neural control tends for kinematic stability while the magnitude of muscle activity varies to a greater extent. To my best knowledge, this is the first study that shows that in arm cranking the kinematic stability is maintained when muscle activation variance is increasing due to altered crank resistance.

I have shown the effect of muscle stimulation patterns on the physical performance of complete spinal cord injured participant in FES-assisted cycling training. I demonstrated the effect of series of training sessions on performance in patients with SCI. Patients achieved a significant increase in performance as a result of the training. The effect of the stimulation on performance is different in the same patient using two different muscle stimulation patterns. If parts of the quadriceps muscle is activated separately (rectus femoris, lateralis vastus and medialis) by coordinated stimulation than greater performance can be achieved. The heart rate and mean arterial pressure measured during 172 training sessions of 10 patients show that due to differences in injury level and type, and general physical, cardiovascular status the patients' performances were different. In general, heart rate, in the average of the 172 trainings, did not increase and then decreased at the end of the training. Mean arterial pressure (MAP) increased and then decreased. The distance and speed increased during mobile tricycling of SCI patients with denervated muscles. To the best of my knowledge, no one has yet studied how the measured parameters change during FES-regulated tricycling with denervated muscles.

My dissertation deals with research on intentionally controlled upper limb cycling and FES-controlled lower limb cycling. This may contribute to the development of exercise protocols to maintain the general physical condition of spinal cord injured persons. People with paralyzed lower limbs have a sedentary lifestyle. Their lower limb cycling can be induced by FES. My work has contributed to this. Additionally, I examined the regulation of arm cycling from a kinematic and force control aspect and found that arm cycling allows greater variance in muscle activity while maintaining kinematic stability. The results of the dissertation may contribute to planning of training protocols for paraplegic persons, combining voluntary arm cycling with FES controlled lower limb cycling.

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