

# PEDALING PARAMETERS BEHAVIOR ON HEALTHY SUBJECTS: TOWARDS A REHABILITATION INDICATION

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## ABSTRACT

It is of outmost importance to identify the quantitative indicators that characterize the rehabilitation degree of the lower limbs of stroke patients and qualitative indicators of the quality of the movement. As a first step in this direction, a cycling ergometer, used in hospitals and rehabilitation clinics, was modified to provide informations about the force applied in the pedals and the pedal angles. One group of non-pathological subjects performed a set of trials at different workloads and cadence values, to analyze the effect of these variables on force output. An increased workload resulted in the raise of the work performed by each leg, whereas the cadence results were inconclusive. Results suggest that the variation of the workload may be a suitable method to characterize motor impairments.

*Index Terms*— Cycling, hospital, force, rehabilitation

## 1. INTRODUCTION

Stroke is a leading cause of long-term disabilities, such as hemiparesis, inability to walk without assistance and dependence of others in the activities of daily living [1]. In particular, hemiparesis is a motor impairment that affects one side of the body. Postural imbalance or asymmetrical movements between the lower limbs are generally observed in hemiparetic patients [2]. Cycling exercise has been investigated as a stroke rehabilitation method and has demonstrated to be a therapy able to improve the lower limbs function [3–5]. It shares similar kinematic pattern with walking, since both are cyclical; require reciprocal flexion and extension movements of hip, knee and ankle; and present an alternating and coordinated antagonist muscle activation [6]. The range of motion (ROM) in cycling is superior than in walking, therefore pedaling helps maintaining the functional range of motion of the lower limbs required to walking [7]. Moreover, cycling is safe, since balance is not a factor in seated pedaling, thus individuals do not experience too risky postural disturbances related to the initial stages of upright locomotor training [4]. This feature and

the ability to exercise while seated allow cycling to be accessible to patients in different disease phases (acute, sub-acute or chronic) [8].

Besides rehabilitation purposes, cycling ergometers present great interest as an assessment tool [2, 9], providing useful informations about the efficiency of the rehabilitation, and in the assessment of the medical state of the patient (the invalidity level). The evaluation of the force, and consequently the work, produced by each leg is one way to assess the motor impairment in lower limbs. Some studies have evaluated the applied force during cycling exercise with different workloads and speeds, in experienced cyclists [10], subjects with lower limb lesions [11] and stroke patients [12, 13].

The aim of this study is to identify the quantitative indicators, which may be employed in the characterization of the lower limbs in stroke patients. In this study, experiments were performed with non-pathological subjects. The analysis of the collected experimental data allows to identify the characteristic behavior of the most significant parameters typical of healthy subjects. In the mean run, the same parameters will be collected and analyzed on pathological subjects, in order to provide for quantitative indicators of the rehabilitation degree of the lower limbs and qualitative indicators of the quality of the movement. A cycling ergometer was modified to provide useful informations about the user-device interaction. This ergometer is normally used in hospitals and rehabilitation clinics, and it will be used in future tests with stroke patients. Similar studies have been performed in [12, 13], but the cycling device was different, and thus the analysis is needed. Further, the used device is the one normally used in Hospital facilities, which will provide for real and effective quantitative indicators of the rehabilitation degree in these facilities.

## 2. METHODS

### 2.1. Ergometer System

A THERA-live motorized cycle-ergometer (Medica Medizintechnik GmbH, Germany), which allows active and passive modes of training, was used in this study, as shown in Figure 1. In the active mode, the subject does all the effort to

This work has been supported by FCT Fundação para a Ciência e Tecnologia within the Project Scope: PEst-OE/EEI/UI0319/2014.



**Fig. 1.** Ergometer system: (a) side view and (b) front view.

move the legs against a specific workload. This mode is similar to cycling in a standard bicycle. In passive mode, the legs of the subject are driven by the pedals of the device, which are moved by a motor.

## 2.2. Instrumentation and Data Acquisition

To obtain measurements of the force applied in the pedals during the different positions of the crank arm, an independent measurement system was developed.

The angle of the crank arm ( $\theta$ ) was measured through two optical sensors and two optical encoders discs already included in the commercial ergometer, which together provide a resolution of  $4^\circ$ .

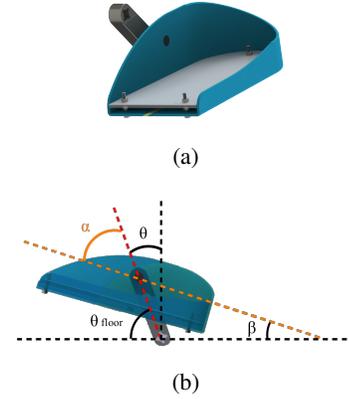
The force applied in the pedals ( $F_p$ ) is measured through one piezoresistive force sensor (Flexiforce sensor) placed on the surface of the pedal, on the line where the crank arm connects to the pedal. The user applies force on the entire surface of the pedal, however, the sensing area of the sensor is only 9.5 mm of diameter. Therefore, a mechanical system was developed to overcome this problem. The system, as shown in Figure 2, is comprised of one metal plate and one acrylic plate with the shape of the pedal, together with a rubber cube, smaller than the sensing area. This cube was placed above the sensing area, to ensure that the load applied by the foot on the plates is transferred to the sensing area through the rubber cube. To avoid force detection when the pedal is unloaded, and to allow the down movement and the stability of the mechanical system, three metal springs and screws were placed between the plates and pedal. Different known weights were used to calibrate the force sensors.

As a pedal moves freely around the crank axis, in order to calculate the effective force, the angle between the crank arm and the pedals ( $\alpha$ ) has to be measured (Figure 2(b)). This angle was obtained through the crank and pedal angles relative to the floor. The angle of the crank arm relative to the floor ( $\theta_{floor}$ ) was calculated as given by (1)

$$\theta_{floor} = 90^\circ - \theta \quad (1)$$

where  $\theta$  is the angle of the crank arm, which is known from the information given by the optical encoders.

An accelerometer (MPU-6050 sensor) was placed under the pedal, in order to measure its angle relative to the floor ( $\beta$ ), where the crank arm connects to the pedal. Finally, the angle



**Fig. 2.** Tridimensional model of the system to measure the applied force. Representation of (a) the mechanical system and (b) the axis and angles to calculate the angle between the pedal and the crank axis ( $\alpha$ ).

between the crank arm and the pedals ( $\alpha$ ) was calculated as follows:

$$\alpha = \theta_{floor} - \beta \quad (2)$$

The effective force ( $F$ ) (force applied at the end of the crank arm), which contributes to the movement of the crank axis, was obtained by (3)

$$F = F_p \times \cos \alpha \quad (3)$$

The value of the torque ( $\tau$ ) was calculated by the product of the effective force and distance, in meters, between the cranks axis and the pedal axis (where the crank arm connects to the pedal) according to (4):

$$\tau = F \times d \quad (4)$$

Finally, the net mechanical work ( $W$ ) applied by the leg can be computed as follows:

$$W = \int \tau d\theta \quad (5)$$

The net mechanical work done by each leg represents the contribution of leg to the movement. It can be positive, meaning that the leg assists crank propulsion; negative, if the leg resists crank propulsion, or zero if the contribution and the resistance to the movement are equal.

## 2.3. Procedure

Three male healthy subjects ( $22.67 \pm 1.15$  years;  $1.80 \pm 0.05$  m;  $78.33 \pm 9.61$  Kg and all with preferred right foot) were recruited to test the developed system. All subjects were asked to perform 9 workload and cadence combinations (workloads: 2, 8 and 15 Nm; cadences: 20, 40 and 60 rpm). The subjects seated in a chair with 43 cm of height. To standardize the condition under which the cycling test was conducted, the

Subject		2 Nm		8 Nm		15 Nm	
		$W_R$	$W_L$	$W_R$	$W_L$	$W_R$	$W_L$
1	20 rpm	-12.16	-3.36	88.61	39.60	275.28	88.23
		(10.38)	(3.07)	(9.59)	(7.50)	(32.62)	(14.70)
	40 rpm	-2.07	7.18	61.71	48.70	132.85	49.08
		(8.25)	(4.05)	(23.21)	(10.62)	(38.26)	(6.63)
	60 rpm	49.60	34.81	52.88	32.63	57.39	35.97
		(3.87)	(6.05)	(11.86)	(11.01)	(5.38)	(3.47)
2	20 rpm	-5.24	3.24	49.50	25.88	92.62	65.87
		(13.24)	(3.10)	(13.65)	(7.76)	(5.93)	(10.47)
	40 rpm	14.32	6.02	56.94	85.00	127.36	98.22
		(4.69)	(7.60)	(16.91)	(23.14)	(7.53)	(12.04)
	60 rpm	62.98	18.06	106.39	45.97	216.72	64.57
		(18.28)	(7.98)	(9.09)	(11.26)	(95.75)	(6.61)
3	20 rpm	1.09	-0.88	87.58	40.28	167.87	108.31
		(0.62)	(3.31)	(11.45)	(14.87)	(23.79)	(15.33)
	40 rpm	10.24	7.13	157.88	72.40	256.69	133.57
		(10.76)	(1.88)	(4.20)	(3.90)	(9.36)	(9.69)
	60 rpm	182.73	42.64	314.70	88.60	393.88	121.48
		(8.43)	(5.85)	(38.06)	(5.20)	(31.27)	(4.40)

**Table 1.** Mean (SD) values of the net mechanical work produced by the right and left limbs ( $W_R$  and  $W_L$ , respectively) for each workload (2, 8 and 15 Nm) and cadence (20, 40 and 60 rpm) combination, for each subject.

distance from the seat edge to the crank axis was adjusted so that the knee extension angle was  $10^\circ$  less from full extension when the subjects extended their knees maximally, as used by Fujiwara *et al.* in [14]. The zero degree of the crank angle ( $\theta$ ) was defined as when the crank arm was perpendicular to the ground surface and the right pedal was on top. To reduce the influence of the weight and inertial forces on the force results, a constant value, computed through a passive exercise of 3 revolutions at 10 rpm, was subtracted from the force values.

The angles values ( $\alpha$  and  $\theta$ ) and the force applied in the force sensors were used to obtain the effective force values, which allow the calculation of the net mechanical work done by each leg. The results were obtained in steady conditions, during one minute for each combination, to ensure a lower deviation on the results.

A two-way analysis of variance (ANOVA) with replication was used to identify the difference of the net mechanical work data for each workload and cadence condition and the interaction between the two variables, for each limb, within each subject. The effects of the factors and interaction were considered statistically significant when the  $p$ -value was  $< 0.05$ .

### 3. RESULTS

The mean values of the pedal angle ( $\beta$ ) and the angle between the pedal and the crank arm ( $\alpha$ ), for five revolutions ( $11^{th}$  to  $15^{th}$ ) in steady conditions, for the three different workloads at 40 rpm, for the right leg of subject 1, with respect to crank

Subject		Workload	Cadence	Workload $\times$ Cadence
1	$W_R$	$<0.01$	$<0.01$	$<0.01$
	$W_L$	$<0.01$	<b>0.047</b>	$<0.01$
2	$W_R$	$<0.01$	$<0.01$	0.232
	$W_L$	$<0.01$	$<0.01$	$<0.01$
3	$W_R$	$<0.01$	$<0.01$	$<0.01$
	$W_L$	$<0.01$	$<0.01$	$<0.01$

**Table 2.** Significance-values resulted from two-way analysis of variance (ANOVA) with replication on workload, cadence and the interaction between these two factors, for each limb, within each subject.

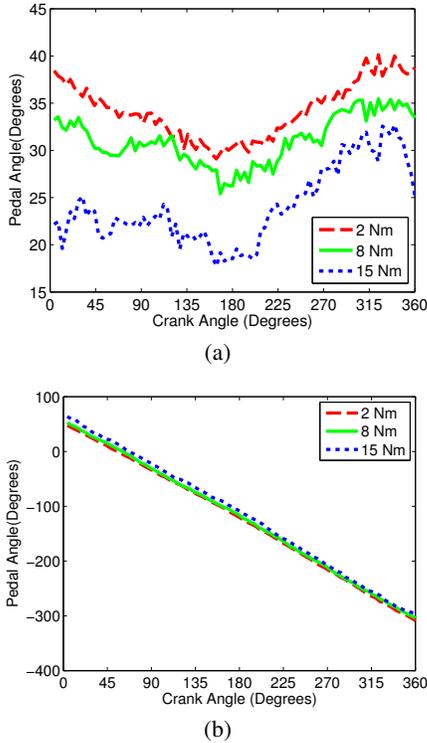
angle are presented in Figure 3 (a) and (b), respectively.

The net mechanical work values produced by the right ( $W_R$ ) and the left ( $W_L$ ) legs for each workload and cadence combination, in the different subjects, are presented in Table 1. In general, the effective positive force increased with the increasing of the workload, whereas the effective negative force decreased. The results obtained with the variation of the cadence are inconclusive, since different trends are observed, for different conditions and subjects. The  $p$ -values for each variable (workload and cadence) and for the interaction between the two variables are presented in Table 2. The effects of workload, cadence and interaction between the workload and the cadence are all statistically significant, except for the interaction in right leg, in subject 2.

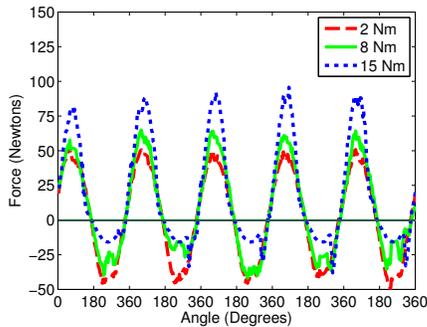
The effective force plotted as a function of changes in workload, throughout five revolutions ( $11^{th}$  to  $15^{th}$ ) in steady conditions, with respect to crank angle, obtained in right leg of subject 2, is presented in Figure 4. The plot for the same variable, as a function of changes in cadence, throughout five revolutions ( $11^{th}$  to  $15^{th}$ ) in steady conditions, with respect to crank angle, obtained in right leg of subject 1, is presented in Figure 5. The recorded signals are very similar to those presented in the literature for similar conditions of cycling [10, 15]. Therefore, it was reasonable to take into account these signals for further analysis.

### 4. DISCUSSION

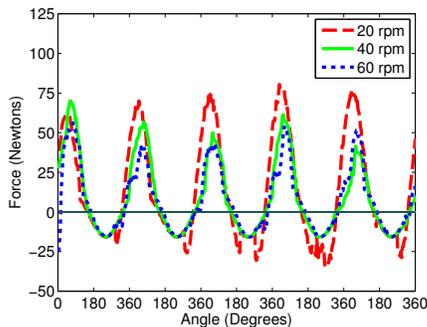
The values of the pedal angle (Figure 3 (a)) for workload tests range between  $17^\circ$  and  $40^\circ$ . The lower values occur at around  $180^\circ$ , which is when the cycling phase transition occurs (downstroke to upstroke, for the right limb, and upstroke to downstroke, for the left limb). As the pedal angle is never zero, the crank angles, for which the effective force is maximal, did not occur precisely at  $90^\circ$  and  $270^\circ$ , for each revolution. In addition, the angles, where the effective force is zero, did not occur precisely at  $180^\circ$  and  $360^\circ$ , for each revolution, for the right pedal. The effect of the workload in the pedal angle is minimal. The  $\alpha$  angle (Figure 3 (b)) shows a linear trend, which indicates that the variability of the pedal angle does not influence significantly the  $\alpha$  angle values. The change of the workload does not have much impact on the  $\alpha$



**Fig. 3.** Mean values of the (a) pedal angle ( $\beta$ ) and the (b) angle between the pedal and the crank arm ( $\alpha$ ), for five revolutions ( $11^{th}$  to  $15^{th}$ ), for the three different workloads at 40 rpm, for the right leg of the subject 1. The maximal values for standard deviation were 2.66 (2 Nm), 3.85 (8 Nm) and 5.27 (15 Nm).



**Fig. 4.** Effective force values, throughout 5 revolutions ( $11^{th}$  to  $15^{th}$  revolution), obtained by the right leg of subject 2, at three different workloads (2, 8 and 15 Nm) and 40 rpm.



**Fig. 5.** Effective force values, throughout 5 revolutions ( $11^{th}$  to  $15^{th}$  revolution), obtained by the right leg of subject 1, at three different cadences (20, 40 and 60 rpm) and 8 Nm.

angle, since the values are almost equal. The plots of (Figure 3 (b)) show that the point, where the effective force is equal to the applied force ( $\alpha$  angle equal to zero), occurs approximately at a crank angle range of  $56^\circ$  to  $68^\circ$ . Additionally, the point where the forces are opposed ( $\alpha$  angle equal to  $270^\circ$ ) occurs approximately between the  $320^\circ$  to  $328^\circ$ .

As the workload increased, the subjects showed an increased net mechanical work done by each leg. This is shown in Table 1, where, for each leg, the value of the net mechanical work increased for higher workloads. This increase results from a combination of a higher positive effective force and a reduction in the negative force, as shown in Figure 4. In some cases, only one response was observed. At higher workloads, it is required to apply more force during downstroke phase ( $0^\circ$  to  $180^\circ$  for the right limb) to overcome resistance and to move the crank, which results in a higher positive effective force. The decrease in the negative effective forces, during upstroke phase ( $180^\circ$  to  $360^\circ$  for the right limb), suggests that the subjects adopted a strategy to improve the effective application of force by reducing the retarding force and, consequently, reducing the need of a higher demand on the propulsive leg, which is in the downstroke phase, to overcome the recovery leg. These results are consistent with previous studies by Sanderson *et al.* [10].

The cadence results are inconsistent amongst the subjects, since different responses were observed under the same conditions. With the increase of cadence, the responses of the subjects and of each leg were different. A trend was observed in few cases, as the one shown in Figure 5. In this situation, a lowered cadence resulted in a higher positive force. This result is consistent with the study of Sanderson *et al.* [10]. However, the authors also report a reduction of the negative force, which in this case, is practically inexistent. In addition to this trend, some other cases show the opposite behavior, an increasing of the effective forces amplitude with the increased cadence. The absence of a trend may result from the effects of the non-muscular components of the applied force, reported by Kautz *et al.* [15]. The authors decomposed the force applied during cycling in two different types: the muscular and the non-muscular force components. A higher cadence resulted on the same trend for muscular and non-muscular force components, although the last one is predominantly responsible for the changes in the applied pedal force. The non-muscular force is comprised of the weight and inertial forces. The authors observed that the first had a lower variance with the cadence. However, the amplitude of the inertial force component increased with the raise of the cadence (higher forces in absolute value) [15]. The weight component was obviated through the subtraction of the measured forces during passive exercise, thus this component may not have considerable impact on the results. As the passive test was performed at 10 rpm, the inertial components may influence the obtained force values, since the tests were performed at higher cadences. The subjects reported a higher

difficulty to maintain a constant cadence for 20 rpm, since they felt a higher resistance. Moreover, the device operation induces some propulsive power, even when the user is pedaling actively. These aspects may result in abnormal changes in the force output.

The high variability of the results between subjects and in different conditions may result of different pedaling strategies adopted by each subject to overcome different workloads and maintain a constant cadence.

## 5. CONCLUSIONS

The variation of the workload produced more consistent results amongst subjects, whereas the cadence variation resulted in distinct patterns. Since the future objective is to characterize the motor function of the lower limbs in stroke patients, a different cadence analysis may not be appropriated to characterize it, due to the effect of the non-muscular force components, which may influence the results and be inconclusive.

Analyzing the influences of these variables in cycling symmetry, as well as, providing feedback about the work values in order to enhance cycling symmetry might be a feasible approach for future studies with stroke patients, since asymmetrical movements are generally observed in these patients.

## REFERENCES

- [1] Alan S Go, Dariush Mozaffarian, Véronique L Roger, Emelia J Benjamin, Jarett D Berry, Michael J Blaha, et al., "Heart Disease and Stroke Statistics—2014 Update: A Report From the American Heart Association.," *Circulation*, Dec. 2013.
- [2] Hsin-Yung Chen, Shih-Ching Chen, Jia-Jin Jason Chen, Li-Lan Fu, and Yu Lin Wang, "Kinesiological and kinematical analysis for stroke subjects with asymmetrical cycling movement patterns.," *Journal of electromyography and kinesiology : official journal of the International Society of Electrophysiological Kinesiology*, vol. 15, no. 6, pp. 587–95, Dec. 2005.
- [3] Michal Katz-Leurer, Iris Sender, Ofer Keren, and Zeevi Dvir, "The influence of early cycling training on balance in stroke patients at the subacute stage. Results of a preliminary trial.," *Clinical rehabilitation*, vol. 20, no. 5, pp. 398–405, May 2006.
- [4] David A Brown, Sabina Nagpal, and Sam Chi, "Limbed cycling program for locomotor intervention following stroke.," *Physical therapy*, vol. 85, no. 2, pp. 159–68, Feb. 2005.
- [5] A Kamps and K Schüle, "Cyclic movement training of the lower limb in stroke rehabilitation," *Neurologie & Rehabilitation*, vol. 5, pp. S1–S12, 2005.
- [6] C C Raasch and F E Zajac, "Locomotor strategy for pedaling: muscle groups and biomechanical functions.," *Journal of neurophysiology*, vol. 82, no. 2, pp. 515–25, Aug. 1999.
- [7] Sang-I Lin, Chao-Chen Lo, Pei-Yi Lin, and Jia-Jin J Chen, "Biomechanical assessments of the effect of visual feedback on cycling for patients with stroke.," *Journal of electromyography and kinesiology : official journal of the International Society of Electrophysiological Kinesiology*, vol. 22, no. 4, pp. 582–8, Aug. 2012.
- [8] Kathryn M Sibley, Ada Tang, Dina Brooks, David A Brown, and William E McIlroy, "Feasibility of adapted aerobic cycle ergometry tasks to encourage paretic limb use after stroke: a case series.," *Journal of neurologic physical therapy : JNPT*, vol. 32, no. 2, pp. 80–7, June 2008.
- [9] J C Rosecrance and C A Giuliani, "Kinematic analysis of lower-limb movement during ergometer pedaling in hemiplegic and nonhemiplegic subjects.," *Physical therapy*, vol. 71, no. 4, pp. 334–43, Apr. 1991.
- [10] D J Sanderson, E M Hennig, and A H Black, "The influence of cadence and power output on force application and in-shoe pressure distribution during cycling by competitive and recreational cyclists.," *Journal of sports sciences*, vol. 18, no. 3, pp. 173–81, Mar. 2000.
- [11] Giovanni Mimmi, Paolo Pennacchi, and Lucia Frosini, "Biomechanical analysis of pedalling for rehabilitation purposes: experimental results on two pathological subjects and comparison with non-pathological findings.," *Computer methods in biomechanics and biomedical engineering*, vol. 7, no. 6, pp. 339–45, Dec. 2004.
- [12] D. A. Brown and S. A. Kautz, "Increased Workload Enhances Force Output During Pedaling Exercise in Persons With Poststroke Hemiplegia," *Stroke*, vol. 29, no. 3, pp. 598–606, Mar. 1998.
- [13] D A Brown and S A Kautz, "Speed-dependent reductions of force output in people with poststroke hemiparesis.," *Physical therapy*, vol. 79, no. 10, pp. 919–30, Oct. 1999.
- [14] Toshiyuki Fujiwara, Meigen Liu, and Naoichi Chino, "Effect of pedaling exercise on the hemiplegic lower limb.," *American journal of physical medicine & rehabilitation / Association of Academic Physiatrists*, vol. 82, no. 5, pp. 357–63, May 2003.
- [15] S A Kautz and M L Hull, "A theoretical basis for interpreting the force applied to the pedal in cycling.," *Journal of biomechanics*, vol. 26, no. 2, pp. 155–65, Feb. 1993.