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Novel solutions for motion analysis

Robotics, clinics and sports applications

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To Prof. Cappa

To Zio Coriolano

I have no special talents,
I am just passionately curious

(A. Einstein)

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Aim of the work

The present PhD thesis is focused on a main topic of the experimental Biomechanics, that is the human motion analysis, and it aims at proposing novel solutions for motion analysis in three main applications, that are rehabilitation robotics, clinics and sports.

Human motion analysis is a general term regarding the automatic description and understanding of human movements using different systems. The rationale of movement analysis is based on the following question: *How is a specific motor activity being performed?* In order to answer this question several outcomes can be taken into account, from kinematic models to muscle activation, including spatiotemporal parameters. Such variables can be gathered by both robotic devices and wearable sensors, enlarging the usual laboratory application scenario to outdoor environments. In addition, the technology supporting the analysis of human motion has advanced dramatically in the last decades, permitting to capture data rapidly, accurately and efficiently. Finally, the use of technology allows to define standards for the experimental procedure and to quantify parameters of interest. From this perspective, movement analysis can be a useful tool in several research fields; specifically, applications in clinics and sports will be discussed in the following chapters.

Then, the main objective of this thesis is to propose novel protocols, algorithms, postprocessing procedures or indices to enhance the potential of human motion analysis performed with robotic devices or wearable sensors in rehabilitation robotics, clinics and sports.

The thesis is divided in two main parts, the first one concerning the robotic devices and the second one focused on the wearable sensors. The first part is further divided in two chapters: (i) the first one aims at proposing a novel procedure to assess the spasticity severity in children with Cerebral Palsy by using a mechatronic robot for the ankle joint; and (ii) the second one at evaluating the effectiveness of a laboratory-made active orthoses for knee and ankle rehabilitation using specific kinematic and spatio-temporal indices. The second part is further divided in four chapters aim respectively at: (i) proposing a novel algorithm for performing automatic gait partitioning in real-time; (ii) proposing a novel procedure to avoid the subject-specific training of a machine-learning algorithm; (iii) proposing a wearable device for the automatic identification of irregularity in race walking; and, (iv) evaluating the repeatability of muscle synergies in order to propose a neurophysiological index based on muscle activity analysis.

SECTION 1

Robotic devices

Introduction

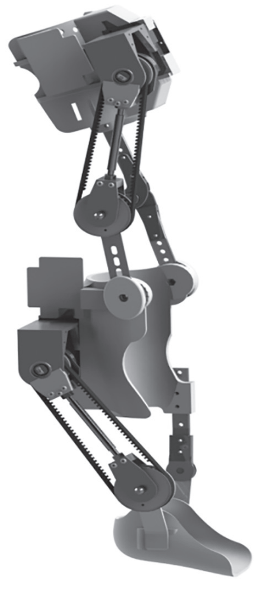
Cerebral Palsy (CP) is one of the most widespread neurological and neuromuscular diseases (NDD) among children population [1,2]. The brain injury in CP is permanent but not progressive and irreversible [1]; thus, the rehabilitation assumes a fundamental role in order to increase the patient quality life in terms of motor ability.

In the last decade, robot mediated therapy became a prominent solution for rehabilitation, as both an alternative and a helping solution to the traditional rehabilitative programs. As regards traditional therapies, the rehabilitation consists in passive handling of the affected limb; it is clear that this type of therapy is strongly related to the physiotherapist's capability and repeatability in terms of speed and force of the imposed movements. Several studies have shown as intensive training focused on performing specific tasks can result in an increase of motor performance related to the affected limbs due to neuronal plasticity that cause a re-organization of motor and sensitive cortex [3]. Consequently, the physical therapies became labor intensive and expensive. Several manual rehabilitation strategies have been proposed to improve autonomy and quality of life in children with neurological disease. From this perspective, a large amount of works have assessed the effectiveness of rehabilitation carried out by robotic devices [3,4]. Actually, robotic devices permit to: (i) customize the rehabilitative therapies based on patient-specific necessity, (ii) perform repetitive and intensive movements; and, (iii) integrate virtual reality with serious games to increase the engagement of patient during rehabilitative section [5]. In addition, robotic devices can represent useful tools not only for the rehabilitation but also as measurement systems. In fact, robotic devices are equipped with several sensors, such as load cell, encoders, inertial measurement units and so on, that can be used to measure kinematic and dynamic variables related to patients' limbs.

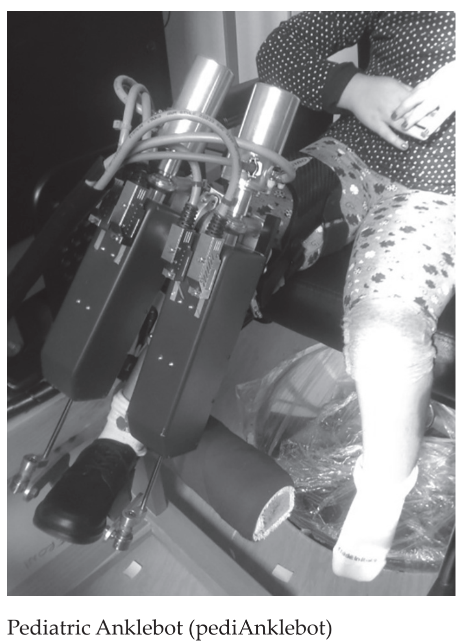
The first part of this thesis is focused on the use of two robotic devices for movement analysis: (i) the pediAnklebot (pediatric ankle robot), a robot addressed to the rehabilitation of the ankle joint in children affected by CP; and (ii) the WAKE-up (Wearable Ankle Knee Exoskeleton), a bi-modular active orthosis for the rehabilitation of the knee and the ankle joint in children affected by CP.

In Chapter 1, the use of pediAnklebot as measurement system will be analyzed to verify the feasibility of evaluating the spasticity severity in children with CP. By answering to this question, we will be able to propose a viable tool for providing objectivity to the spasticity assessment currently performed by means of clinical scales that are operator-dependent.

In Chapter 2, the use of WAKE-up as a robotic device to assist gait in children with CP will be showed. The effectiveness of the orthoses will be evaluated considering the benefits in terms of gait kinematic, gait variability and gait asymmetry.



Wearable Ankle Knee Exoskeleton (WAKE up)



Pediatric Anklebot (pediAnklebot)

Chapter 1
Spasticity assessment by means of pediAnklebot[1]

1.1. Introduction

Cerebral Palsy (CP) is characterized by different forms of hypertonia with symptoms defined by three main descriptive terms: spasticity, dystonia, and rigidity [6–8]. The spasticity, in turn, is commonly defined as a hyperexcitability of the stretch reflex responses, which are velocity-dependent. More specifically, spasticity can be identified as an increasing resistance with the increase of the joint passive rotation or as a rapid rise of the resistance above a threshold speed or joint angle value [7,9,10].

In clinical setting, spasticity is usually assessed through the Modified Ashworth Scale (MAS) [11]. However, its validity and reliability as a measure of spasticity has been questioned [12,13]. In addition, no significant correlation was found between MAS and measurements of neural and muscular components of joint static and dynamic stiffness [14]. The main limit of the spasticity assessment using MAS is the impossibility to distinguish between neural and non-neural components. In fact, the mechanical resistance perceived by the examiner is not only addressable to the active muscle response against induced elongation, but also to the elastic and viscous characteristics of muscles, tendons, soft tissues and ligaments. MAS does not take into account the speed dependence of the stretches in induced passive movements, a potential

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1 Germanotta M.*; Taborri J.*; Rossi S.; Frascarelli F.; Palermo E.; Cappa P.; Castelli E.; Petrarca M. Spasticity measurement based on tonic stretch reflex threshold in children with Cerebral Palsy using pediAnklebot. *Frontiers in Human Neuroscience* 2017, 11(277).
* These authors contributed equally to this work

effective parameter for assessing the neural component of spasticity [15]. Consequently, MAS can be effectively used as an index of the resistance to passive movements rather than spasticity [13]. Among the clinical scales for spasticity assessment, the Tardieu Scale represents a potential alternative to the MAS scale, as it also considers velocity-dependent characteristic of stretch reflex [16,17]. However, it is affected by operator's skills in administration, i.e. the ability to manually impose angular velocity, and by the difficulty to induce fast velocities in children [18]. Thus, its reliability is still questioned [19].

Even though clinical scales still represent an integral part of neurological and clinical examination, it is widely acknowledged that instrumented approaches are desirable, to overcome the intrinsic limit of the clinical approach based on 'feel the resistance'. Several objective approaches based on the analysis of the exerted force and the muscle activity were proposed to assess spasticity [20–24]. In some studies, the spastic limb was manually moved by an expert operator, while the angular displacement and velocity were measured by sensors, such as electrogoniometers [25] or inertial measurement units [26], placed on the subjects. In others, the targeted limb was moved around the anatomical joint by mechatronic devices [23] to standardize and automate the imposed input and improve the test-retest reliability. With this approach, joint displacements, exerted forces, and muscle activities are measured via sensor systems embedded in mechatronic devices [26,27]. Such quantitative and robust measurements could become a valuable tool to reliably monitor patients' progress and assess efficacy of current clinical approaches in spasticity management.

Among the proposed metrics based on sensor outputs, the Tonic Stretch Reflex Threshold (TSRT) index seems to be the more promising approach [28,29], as it more accurately reflects the Lance's definition of spasticity [10] than other clinical tests [15]. In particular, TSRT is estimated by a linear regression model on the Dynamic Stretch Reflex Threshold (DSRT) values, corresponding to the points on a phase diagram (joint velocity vs. joint angle) at which motoneurons and relative muscles begin to be recruited [28]. Levin *et al.* [29,30] demonstrated the correlation between TSRT and the degree of spasticity at the elbow joint in adult population. Levin and colleagues also verified that TSRT angle lies within the biomechanical range of motion (ROM) of the joint in subjects with spasticity, while it is outside those limits in healthy subjects. The TSRT value and the slope μ of the regression line were effectively used to evaluate the hyperton at the elbow joint, both in patients with stroke [28] and CP [31], showing moderate to good values of reliability in both cases. TSRT and μ values were also used to discriminate between neurological deficits of muscle tone at the elbow joint in patients with stroke and Parkinson's disease [29]. Recently, Blanchette *at al.* [32] extended the TSRT approach to the ankle joint as it is one of the most affected joint in patients with spasticity and it significantly affects gait [33]. Blanchette and colleagues found a high inter-evaluator reliability (ICC= 0.85) for plantarflexor spasticity in adult post stroke patients.

As regards children with CP, ankle spasticity was extensively studied using several approaches [21,23,26,34] based on the manual imposition of passive displacements of the foot and the analysis of the SEMG signal. Even though some researches [35,36] proposed the evaluation of spasticity in CP population using a mechatronic device, no studies, to the authors' knowledge, examined the spasticity of plantarflexor and dorsiflexor muscles in pediatric population by means of a mechatronic device, applying the TSRT approach.

Hence, the aim of the present work is twofold. Firstly, we seek to evaluate if TSRT approach can be adopted for the evaluation of the ankle spasticity in children with CP, considering both plantarflexor and dorsiflexor muscles. Secondly, we want to assess if there is a correlation between MAS and quantitative measures obtained from the TSRT approach. For both aims, we used the pediAnklebot (InMotion Technologies, Watertown, MA, USA) [37,38], a robotic device, to impose passive stretches to the ankle joint measuring both rotation and angular velocity.

1.2. Materials and Methods

1.2.1. Participants

A cohort of ten patients was recruited (6.4 ± 1.8 years, range: 5-9), eight children with hemiplegia (HC) and two with diplegia (DC). Inclusion criteria for patients were: (i) congenital or acquired hemiplegia or diplegia, excluding those with acute events in the last six months; (ii) age between 5 and 10 years; (iii) no cognitive or visual impairments; (iv) no botulinum toxin injection prior six months; and, (v) no history of functional surgery. A cohort of three age matched [39] typically developing children (TDC) (6.3 ± 0.6 years, range 6-7) was enrolled. The subjects met the following inclusion criteria: absence of neurological or musculoskeletal disorders, long term medications, bone lesions or joint pathologies of the lower limbs in the year prior to the study. TDC were enrolled to verify the robustness of the algorithm to false positives, i.e. the erroneous individuation of spasticity level in healthy subjects. More details of the two groups and the MAS assessment of the Gastrocnemius are reported in the Tab. 1.

Tab. 1. Participants of the two groups involved in the study. HC stands for Children with Hemiplegia, DC stands for Children with Diplegia, TDC stands for Typically Developing Children, and MAS stands for Modified Ashworth Scale.

	ID	Age (year)	Height (mm)	Body mass (kg)	Diagnosis	Passive ROM (°) Right side	Passive ROM (°) Left side	MAS (Gastrocnemi)
HC	#1	5	1195	22	Right hemiplegia	66	69	2
	#2	5	1180	23	Left hemiplegia	67	64	3
	#3	5	1190	26	Right hemiplegia	69	71	1
	#4	5	1220	27	Left hemiplegia	63	59	4
	#5	8	1270	30	Right hemiplegia	62	65	3
	#6	5	1190	23	Right hemiplegia	68	71	1+
	#7	9	1290	32	Right hemiplegia	65	70	2
	#8	9	1300	33	Right hemiplegia	67	69	1+
DC	#9	5	1250	25	Diplegia	64	62	2
	#10	8	1310	34	Diplegia	60	59	4
TDC	#1	6	1280	32	Healthy	70	-	-
	#2	7	1275	31	Healthy	71	-	-
	#3	6	1290	33	Healthy	73	-	-

All subjects were naive to the robotic device and the task. Written informed consent was obtained from all parents or legal guardians of the children involved in the study. The Research Ethics and Medical Board of the Bambino Gesù Children's Hospital approved the experimental protocol, compliant with the ethical standards outlined in the 1964 Declaration of Helsinki. Experimental trials were performed at the Movement Analysis and Robotic Laboratory (MARLab) at Bambino Gesù Children's Hospital, where patient were recruited.

1.2.2. Experimental set-up

The pediAnklebot [38,40] imposes torques and rotations with programmed angular velocities to the ankle joint and, at the same time, measures their values. In this study, the pediAnklebot was used to impose stretch movements, i.e. rotations at different angular velocities to the ankle joint. It is a backdriveable device with low-friction and low mechanical impedance, permitting normal ROM of the foot with respect to the shank in all anatomical planes. Two linear actuators mounted in parallel compose the kinematic design of the mechatronic device. When the actuators push or pull in the same direction a dorsi-plantarflexion torque is produced at the ankle joint, with a maximum torque supplied of 7.21 Nm. Rotations of the ankle in the sagittal plane can be imposed with a ROM from 65° to 135°. The pediAnklebot is actuated by two brushless DC motors (Maxon EC-powermax 22-327739). Two mini-rail linear encoders (MNS9-135 length, Schneeberg-

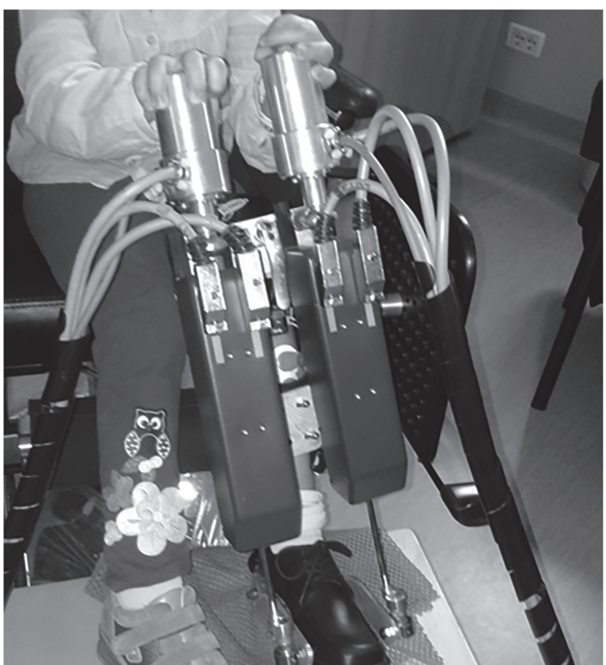

Fig. 1. - A children with hemiplegia wearing the PediAnklebot and instrumented with SEMG.

er), mounted in parallel to the actuators and possessing a resolution of 1 μm, are used to estimate ankle angles in dorsi-plantarflexion. The outputs of two rotary encoders with 40,960 lines are gathered for the servo-amplifier commutation.

For measuring muscle activities, passive surface Ag/AgCl circular electrodes (BlueSensor M, Ambu, Ballerup, Denmark) and a wireless SEMG system (Wave, Cometa, Milan, Italy) were applied on tibialis anterior (TA) muscle, involved in dorsiflexion, lateral gastrocnemius (LG), and medial gastrocnemius (MG) muscles, both involved in plantarflexion. Electrode positions and inter-electrode distances were selected in accordance to the SENIAM guidelines [41]. The wireless SEMG system guarantees no time delay in the acquired signal during the experimental acquisition. The entire experimental setup is shown in Fig. 1.

1.2.3. Experimental protocol

Before the experimental protocol, spasticity was clinically assessed using the Modified Ashworth Scale (MAS), administered by a skilled physiotherapist. Then, the subject sat comfortably on an adjustable chair and the pediAnklebot was mounted on the lower limb. He/she underwent a familiarization training session, which lasted until participants felt familiar with the equipment. The same physiotherapist

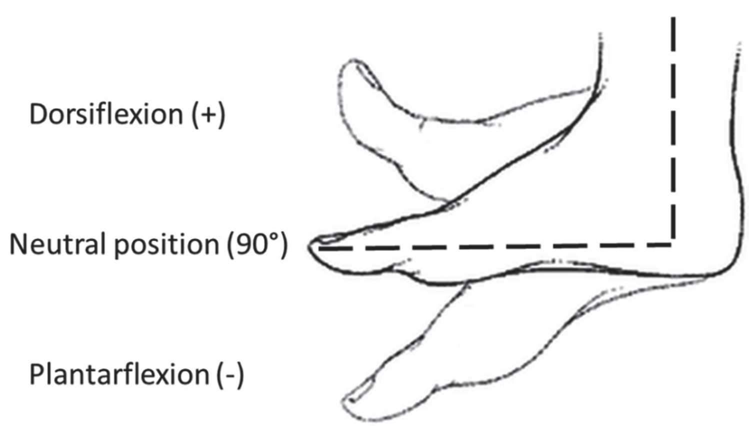

Fig. 2. Definition of joint angles used in the study: neutral position (90°), dorsiflexion (positive values) and plantarflexion (negative values).

placed the ankle in neutral position, i.e. an angle of 90° in sagittal plane between foot and shank [38], while the foot was resting on a support. The neutral position and the direction of dorsi-plantarflexion were reported in Fig. 2.

This position was maintained for ten seconds to record the baseline muscle activity from the SEMG outputs. Passive ROMs of the ankle joint both in dorsi and plantarflexion were then recorded for each subject. Then, the pediAnklebot imposed passive stretch angles moving the ankle alternatively from the 90% of the dorsiflexion passive ROM to the 90% of the plantarflexion ROM, at five different velocities (50 °/s, 100 °/s, 150 °/s, 200 °/s and 250 °/s). For each combination of direction and velocity, the perturbation was applied five times, making a total of 25 stretch movements in dorsiflexion and 25 stretch movements in plantarflexion. To evaluate the accuracy of the pediAnklebot in imposing angular velocities, two preliminary tests were performed. The first test was conducted without the subject's limb linked to the device and the maximum relative error between the imposed and the measured angular velocity was equal to 2%. The second test was performed with the device mounted on a subject and a maximum relative error of 4% was found.

The trial did not start until the participant indicated that he/she was ready to begin, and muscles were relaxed, as showed by the SEMG signals. For each direction, the velocity sequence was randomized and varied among subjects; this variability was introduced to avoid a "muscle accommodation" effect [29,42]. To limit the influence of the previous applied stretches on the response to the following, the stretches were separated by a minimum time of ten seconds [29]. Moreover, an additional variable time (between 0 and 2 s) was added to the minimum time interval to avoid subject anticipation. Therefore, stretches were interspersed with a time interval randomly varying between ten and twelve seconds. During each trial, subjects were asked to relax their muscles as much as possible, and, in addition, their attention was captured with the projection of their preferred cartoons, to avoid voluntary interaction with the movement imposed by the robotic device. The cartoon's volume was kept higher than the noise of the actuators, thus children were could not perceive the beginning of the motor's movement. The session lasted approximately 45 minutes per participant.

SEMG signals of the three above-mentioned muscles, the angular velocity and the angle time histories were acquired. The entire protocol was automatically performed by an ad-hoc control algorithm implemented into the pediAnklebot, programmed to impose passive stretches with random velocities, and synchronize the SEMG with pediAnklebot outputs.

The protocol was performed by the same operator and was repeated for both sides of the lower limb in children with CP, and only for the dominant side in TDC. Dominant side assessment was done by asking to kick a ball [43].

1.2.4. Data processing and data analysis

The data post-processing was performed using Matlab software (MathWorks, 2012b, Natick, MA, USA). According to the literature [29,31], raw SEMG signal was firstly filtered with a band-pass (20 - 500 Hz) 2nd order zero-phase Butterworth filter and, then, a low-pass 30 Hz 6th order zero-phase Butterworth filter was applied to the rectified signal to extract its envelope [44]. The mean value, addressed as baseline, and standard deviation (SD) were evaluated on the rectified SEMG signals that were acquired with the ankle in neutral position in 100 ms windows before the application of each passive stretch. The stretch reflex onset was defined from the SEMG envelope as the first sustained burst that appeared and remained above the baseline + 3 SDs for at least 15 ms. The joint angles, i.e. the dynamic stretch reflex thresholds (DSRTs), and the angular velocities (ω), corresponding to the stretch reflex onsets, were extracted from data provided by the pediAnklebot. The following linear dependency between DSRT and ω can be assumed as also reported in [29,31]:

$$DSRT = TSRT - \mu\omega$$

The linear regression model was applied both to DSRT and ω, to compute the value of μ and TSRT. The μ value represents the sensitivity of DSRT to velocity and it is defined as the angular coefficient of the regression line. In particular a positive value of μ indicates a damping response, while a negative value shows an anti-damping velocity-dependent response to muscle stretch. TSRT was defined as the angle at which muscles are activated during quasi-static stretching; thus, TSRT was estimated as the intercept of the regression line with the angle axis. According to Jobin and Levin [31], the TSRT values fall into the biomechanical ankle range (from 140° for the plantarflexion to 70° for the dorsiflexion [45] in subjects with spasticity, while they are outside the range in typically developing children. Furthermore, TSRT can be related to spasticity severity. In dorsiflexion movements that elicited stretch reflex in MG and LG, values closer 140° are related to a higher pathology severity. Instead, in plantarflexion movements, which elicited stretch reflex in TA, the spasticity severity increases when the TSRT decreases from 140° to 70°.

To evaluate the quality of the linear regression, we used the correlation coefficient (r). The absolute value of r can be interpreted, in agreement to [46], as: (i) no correlation, if $|r| \leq 0.1$; (ii) mild/modest correlation, if $0.1 < |r| \leq 0.3$; (iii) moderate correlation, if $0.3 < |r| \leq 0.6$; (iv) strong correlation, if $0.6 < |r| < 1$; and, finally, (v) perfect correlation, if $|r| = 1$.

Before the evaluation of TSRT and μ, SEMG signals were accurately screened to avoid false detections of stretch reflex. Specifically, trials were discarded when following cases occurred [47]: (i) simultaneous antagonist muscle activity; (ii) presence of clearly identifiable movement artifacts; and, (iii) continuous muscle activation before or during the imposition of passive stretches. Thus, the linear regression was not performed and the associated metrics was not computed when the number of detected DSRTs was less than 6, in accordance to Ferreira *et al.* [48]. After data reduction, the median value of DSRT referred to the elicited number of stretch reflexes for plantarflexors and dorsiflexor was calculated, by considering separately the less and more affected sides for HC and DC, and the dominant side for the TDC.

A paired t-test on the TSRT values related to MG and LG was performed, considering data gathered from the most affected side of HC and both sides of DC. Test was conducted, with α equal to 0.05, to evaluate if muscles recruited in the same movements show the same behavior in terms of TSRT.

Correlations between TSRT and μ related to the same side were determined using Pearson's coefficient, while correlations between objective indices (TSRT and μ) and MAS were determined using the Spearman's test. A p value lower than 0.05 was chosen to indicate statistical significance for all the comparisons. Statistical power of the Spearman's tests was computed using G*power software [49]. Statistical analysis was conducted with SPSS package (IBM-SPSS Inc., Armonk, NY, USA).

1.3. Results

Fig. 3 shows joint angle, angular velocity, torque and relative EMG signals of the plantarflexor responses for a representative patient.

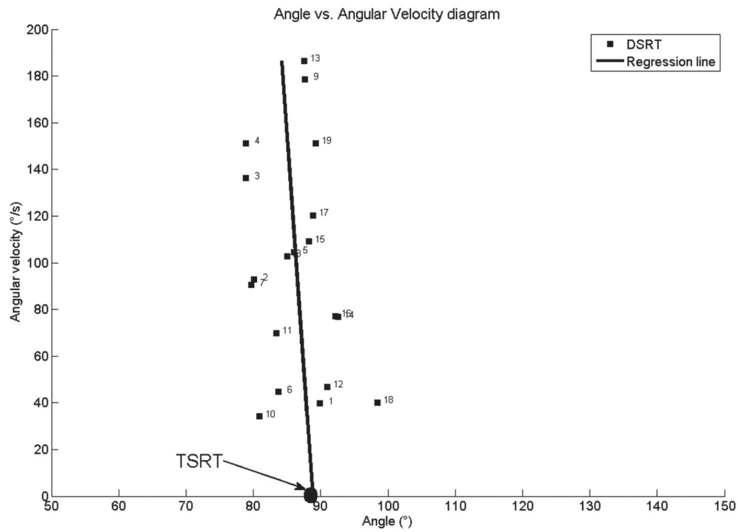

Fig. 3. Patient with hemiplegia #1: Angle vs. Angular velocity diagram and computed parameters for MG.

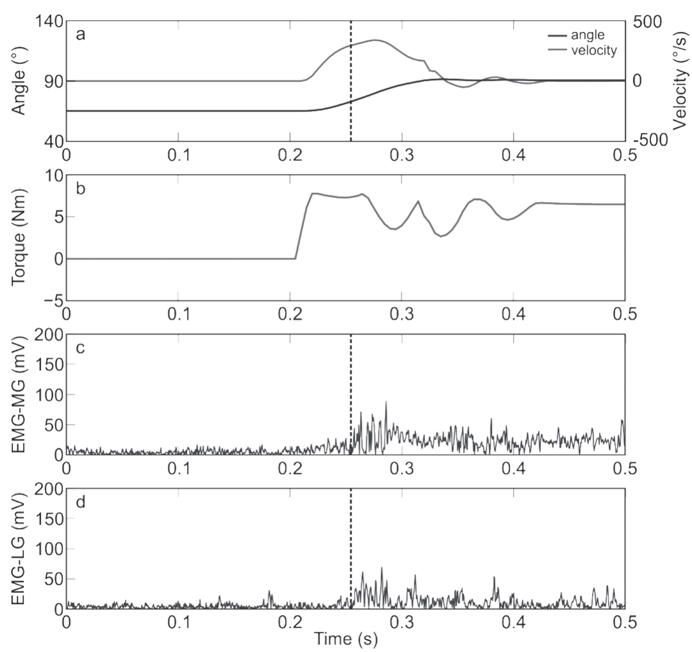

Fig. 4. A representative figure of the imposed angle, angular velocity, torque and relative EMG signals of the Medial and Lateral Gastrocnemius (Patient #10). (a) imposed angle in blue and angular velocity in green; (b) torque; (c) raw data of MG; and (d) raw data of LG. Dotted black line indicates the stretch reflex onset.

A paradigmatic relationship between angular velocity and joint angle, along with the regression line and the related parameters is reported in Fig. 4.

The values of Dynamic Stretch Reflex Threshold DSRT, Tonic Stretch Reflex Threshold TSRT, the sensitivity to stretch reflex μ and the coefficient of correlation r relative to the three examined muscles for all subjects are reported in Tab. 2.

Tab. 2. DSRT, TSRT and μ of the line interpolating velocities with DSRT values for all subjects relative to MG, LG and TA. ND means "not-definable", * represents the negative angular coefficient of regression line and # represents TSRT values outside the healthy biomechanical ROM. MA indicates the most affected side and LA the less affected one.

		MG				LG				TA			
	Participants	**N° DSRT**	**TSRT (°)**	**μ (s)**	**r**	**N° DSRT**	**TSRT (°)**	**μ (s)**	**r**	**N° DSRT**	**TSRT (°)**	**μ (s)**	**r**
HC	#1 MA	20	84.2	0.062	-0.52	20	90.4	0.034	-0.31	4	ND	ND	ND
	#2 MA	15	99.2	0.008	-0.33	16	97.6	0.016	-0.30	5	ND	ND	ND
	#3 MA	12	77.3	0.123	-0.82	17	90.3	0.059	-0.36	8	71.1	0.026	-0.68
	#4 MA	14	94.5	0.082	-0.32	14	81.3	0.094	-0.33	4	ND	ND	ND
	#5 MA	16	101.0	-0.050*	0.32	6	70.1	0.105	-0.75	5	ND	ND	ND
	#6 MA	10	96.3	0.057	-0.34	15	103.5	0.051	-0.44	20	61.7#	0.051	-0.92
	#7 MA	14	126.2	0.014	-0.20	14	112.9	0.107	-0.47	19	91.0	0.026	-0.56
	#8 MA	13	93.0	0.046	-0.49	9	70.2	0.200	-0.84	13	80.0	0.035	-0.71
DC	#9 MA	13	88.1	0.109	-0.45	13	55.6#	0.252	-0.58	0	ND	ND	ND
	#10 MA	15	97.2	0.027	-0.32	14	95.1	0.029	-0.30	1	ND	ND	ND
HC	#1 LA	11	105.3	-0.048*	0.43	4	ND	ND	ND	6	77.9	0.012	-0.52
	#2 LA	8	91.8	0.022	-0.48	9	104.4	-0.026*	0.27	5	ND	ND	ND
	#3 LA	3	ND	ND	ND	5	ND	ND	ND	10	74.7	0.022	-0.37
	#4 LA	12	94.1	0.090	-0.39	12	89	0.560	-0.32	9	114.0	-0.142*	0.56
	#5 LA	3	ND	ND	ND	3	ND	ND	ND	5	ND	ND	ND
	#6 LA	3	ND	ND	ND	8	126.5	-0.062*	0.30	9	100.0	-0.063*	0.59
	#7 LA	3	ND	ND	ND	5	ND	ND	ND	5	ND	ND	ND
	#8 LA	10	77.3	0.141	-0.71	10	114.1	-0.155*	0.41	5	ND	ND	ND
DC	#9 LA	10	82.7	0.162	-0.82	10	87.5	0.135	-0.57	11	84.5	-0.380*	0.45
	#10 LA	10	90.8	0.089	-0.61	10	99.6	0.042	-0.30	1	ND	ND	ND
TDC	#1	9	49.1#	0.183	-0.51	9	65.2#	0.009	-0.36	11	53.1#	0.021	-0.71
	#2	9	56.8#	0.117	-0.62	8	69.6#	0.114	-0.35	11	59.4#	0.025	-0.51
	#3	8	42.0#	0.242	-0.60	10	67.7#	0.106	-0.52	1	ND	ND	ND

r range	no	modest	strong	perfect

1.3.1. Dynamic Stretch Reflex Threshold (DSRT)

As regards the most affected side of children with CP, passive stretches were able to elicit the stretch reflex of the muscles involved in the plantarflexion, i.e. MG and LG, as demonstrated by the number of DSRT, which resulted never lower than six and its median was equal to 14. Conversely, for the muscle involved in the dorsiflexion, TA, the number of DSRT was greater than 6 only in four cases and its median was equal to 16. Furthermore, the number of elicited DSRT was lower in the less affected side than in the most affected one. In particular the number of DSRT was greater than 6 in seven (median equal to 10) and four (median equal to 9) cases for plantarflexor and dorsiflexor muscles, respectively. In typically developing children, the stretch reflex was elicited except for TA and for one subject. The median was equal to 9 for MG and LG and 11 for TA.

1.3.2. Tonic Stretch Reflex Threshold (TSRT)

In the most affected side of children with hemiplegia and diplegia, all computed TSRTs were into the biomechanical ROM, with the exception of those of patients #6 and #9 for the TA and LG, respectively. More specifically, the mean and the SD of TSRT values were 95.7° (12.9°) and 86.7° (17.4°) respectively for MG and LG, and 75.9° (12.5°) for TA. In the less affected side, no patient showed a TSRT value outside the biomechanical ROM. The mean and the SD of TSRT values were 90.3° (9.7°) and 103.5° (14.9°) for MG and LG, respectively, and 90.2 ° (16.5°) for TA. In TDC group, TSRT values were always outside the biomechanical range both in dorsiflexion and in plantarflexion; the mean and the SD of TSRT values were 49.3°(7.4°) and 67.5°(2.2°) respectively for MG and LG, and 56.2°(4.4°) for TA.

1.3.3. Sensitivity to stretch reflex (μ)

Positive values of sensitivity to stretch reflex μ were found in 82% of the computed linear regressions. Negative values of μ occurred in eight cases, one for the most affected side and seven for the less affected one. Specifically, for plantarflexor muscles of the most affected side, the mean value of μ was 0.078 and only subject #5 showed an inverse sensitivity to stretch reflex ($\mu = -0.050$) related to MG. For dorsiflexor muscle, the mean μ value was equal to 0.034. In the less affected side, 67% (mean $\mu = 0.155$) and 40% (mean $\mu = 0.017$) of positive sensitivity values were found for plantarflexor and dorsiflexor muscles, respectively. In TDC, no subject showed negative values of μ. In particular, we found mean values of μ equal to 0.128 both in MG and LG, and 0.023 in TA.

1.3.4. Correlation coefficient r

Considering the plantarflexor muscles of the most affected side, the r values lied in the range of moderate to strong correlation, with one exception of no correlation ($r = 0.20$) for MG in patient #7. The observed mean value for r was equal to -0.44, ignoring the positive value obtained for MG in patient #5. As concerns dorsiflexor muscle, moderate to strong correlations were observed for all patients with a mean value equal to -0.72. In the less affected side, moderate to strong negative correlations were found both for plantarflexor and dorsiflexor muscles, with mean values equal to -0.52 and -0.44, respectively. Moreover, seven occurrences of positive, from modest to moderate, correlations were observed, four in MG and LG, and three in TA.

In TDC, all coefficients were in the range from moderate to strong correlation for all muscles and no positive correlation was found.

1.3.5. Correlation MAS vs.TSRT, MAS vs. μ, and μ vs.TSRT

We did not administer the MAS scale for TA, since the spasticity phenomenon mostly affects the plantarflexors [50]. Spearman's and Pearson's coefficients (r) and relative p values are showed in Tab. 3.

Power of the test resulted equal to 70% with a medium effect size (0.5) [51]. All tests should be interpreted as exploratory data analysis. Thus, such a power level can be considered acceptable for the present study.

MAS values did not show any correlation with the objective indices computed for the most affected side. By comparing TSRT and μ values of different muscles, no statistical correlation was found for both lower limb sides. Conversely, a positive strong correlation was found both for MG and LG between TSRT and μ computed for the same muscle on both most affected and less affected sides. A similar effect was not observed in TA.

Tab. 3. Spearman's and Pearson's coefficients r related to the performed correlation and relative (p-value) relative to Medial Gastrocnemius (MG), Lateral Gastrocnemius (LG) and Tibialis Anterior (TA). Empty cells stands for not performed correlation. Spearman's test was used for correlation between MAS and TSRT and μ; while Pearson's test for correlation between TSRT and μ. * indicate significant correlations.

			Clinical score	**MG**		**LG**		**TA**	
	Muscles	Parameters	MAS	TSRT	μ	TSRT	μ	TSRT	μ
Clinical score		MAS		-0.49 (0.15)	-0.41 (0.23)	-0.06 (0.88)	-0.13 (0.72)		
Most affected side	**MG**	TSRT	-0.49 (0.15)		0.70 (0.04)*	0.47 (0.17)	0.01 (0.98)	0.76 (0.24)	0.47 (0.93)
		μ	-0.41 (0.23)	0.70 (0.04)*		0.18 (0.62)	0.24 (0.50)	0.51 (0.50)	0.31 (0.55)
	LG	TSRT	-0.06 (0.88)	0.47 (0.17)	0.18 (0.62)		0.74 (0.01)*	0.45 (0.55)	-0.49 (0.32)
		μ	-0.13 (0.72)	0.01 (0.98)	0.24 (0.50)	0.74 (0.01)*		-0.18 (0.82)	0.80 (0.06)
	TA	TSRT		0.76 (0.24)	0.51 (0.50)	0.45 (0.55)	-0.18 (0.82)		0.74 (0.26)
		μ		0.47 (0.93)	0.31 (0.55)	-0.49 (0.32)	0.80 (0.06)	0.74 (0.26)	
Less affected side	**MG**	TSRT			0.90 (0.01)*	-0.53 (0.27)	-0.56 (0.33)	-0.06 (0.96)	-0.99 (0.07)
		μ		0.90 (0.01)*		-0.17 (0.75)	0.07 (0.91)	-0.24 (0.84)	-0.95 (0.19)
	LG	TSRT		-0.53 (0.27)	-0.17 (0.75)		0.86 (0.03)*	0.22 (0.78)	-0.37 (0.63)
		μ		-0.56 (0.33)	0.07 (0.91)	0.86 (0.03)*		-0.04 (0.97)	-0.64 (0.56)
	TA	TSRT		-0.06 (0.96)	-0.24 (0.84)	0.22 (0.78)	-0.04 (0.97)		0.17 (0.78)
		μ		-0.99 (0.07)	-0.95 (0.19)	-0.37 (0.63)	-0.64 (0.56)	0.17 (0.78)	

1.4. Discussions

The present study explored the feasibility of using a robotic device for the assessment of spasticity at the ankle joint in children with CP, through the TRST approach. The obtained metrics was compared to the MAS clinical scale.

1.4.1. Is the TSRT approach feasible for the spasticity assessment for Medial Gastrocnemius, Lateral Gastrocnemius and Tibialis Anterior?

The proposed experimental protocol has the potential to elicit a high number of stretch reflexes in medial and lateral gastrocnemius muscles. Therefore, it confirms the results obtained by Blanchette *et al.* [32] in adult post-stroke patients on the same joint. The here adopted approach is sensitive to pathology severity, eliciting a higher DSRT number in the most affected side than the less affected one.

Conversely, the assessment of spasticity via TSRT approach for the tibialis anterior cannot be performed due to the low number of gathered DSRT, i.e. lower than the generally accepted threshold value of 6 [48]. The reduced capability in eliciting a high number of stretch reflexes on the tibialis anterior can be ascribed to the absence of the hyperexcitability phenomenon, as it is generally understood that, in children with CP, spasticity mainly affects plantarflexor muscles [35]. The observed low values of DSRT for the TA in the present experimental study cannot be compared with those of previous studies, due to the novelty in both the targeted muscle and the population.

Focusing on TSRT, the entire set of computed values for the three examined muscles of the most affected side lied into the biomechanical ROM, with the only exception of patient #9 in LG and patient #6 in TA. This outcome demonstrates that spasticity is related to a hypertonicity phenomenon, causing a decrease in the range of central regulation of tonic stretch reflex, and the inability to prevent activation of muscle within biomechanical range [29]. The ability to regulate tonic stretch reflex threshold in a range outside the biomechanical ankle ROM is crucial for the movement control [29]. Deficit in TSRT regulation is considered a leading cause of instability, weakness, and impairments in interjoint coordination [52]. Moreover, the regulation of the stretch reflex permits to convert the reflex resistance in the force that assists the motion [53]. Although no statistical difference was found between TSRT in MG and LG ($p=0.12$), we observe that in some patients (HC#3, HC#5, HC#8) values of TSRT were different between MG and LG. This finding suggests that the hyperexcitability hides de-facto a complex interaction between central and peripheral contributions to spasticity. Thus, muscles recruited in the same movement can differently contribute depending on the pathology severity.

The observed differences in TSRT values among patients with the same MAS score might be due to the subject-specific response to the externally imposed joint rotations, as well as by the intrinsic inability of the MAS to discriminate neural from non-neural components. The computed μ values confirm the outcomes described by Blanchette and colleagues [32], who demonstrated a lower sensitivity to the stretch velocity of the ankle plantarflexors than the elbow muscles. This outcome could be ascribed to the higher passive stiffness of the ankle joint with respect to the elbow one [54]. The negative values of μ prove an atypical opposite dependency of DSRT to the velocity. The lower the velocity of the imposed rotations the earlier occurs the stretch response. This effect might be due to the common occurrence of muscle shortening reactions in patients with neuromotor diseases [55]. Moreover, a low sensitivity to stretch reflex suggests a strong presynaptic inhibition of primary fiber of type Ia afferent discharges [29].

The computed correlation coefficients r for plantarflexor (MG and LG) and dorsiflexor (TA) muscles were lower than those obtained by using the same methodology at the elbow joint [29,31] or at the ankle joint for adult post-stroke patients [32]. We speculate that the previously indicated outcomes could be influenced by: (i) the lower ROM value for the ankle compared to the elbow one; (ii) the targeted children population; and (iii) the level of the pathology, which is mild or slight in the recruited patients. The low values of correlation coefficient could also be ascribed to the lower cooperation of children with respect to adults [56], leading to a higher number of discarded trials for the presence of voluntary movements. Similarly, the observed differences in the r values related to the MG and LG in the same patient could be due to the different number of DSRT values used for the computation of the regression line, as also reported by Ferreira *et al.* [48].

In typically developing children, the computed TSRT values confirm that healthy children are able to control stretch responses outside the biomechanical range, providing full extension of the ankle ROM in the sagittal plane. The DSRT number different from zero confirms the findings of Pisano *et al.* [57], who argued the possibility to evoke involuntary muscle responses also in healthy individuals.

The positive correlation between TSRT and μ computed for MG, LG of both sides, confirmed that higher TSRT values imply a higher slope of the regression line, as reported by Jobin and Levin [31]. Conversely, in correspondence of low TSRT values we found low μ values.

This finding suggests that a more stable response to the stretch reflex is reached. The absence of significant correlations between TSRT and μ in TA suggests that this relationship only exists in muscles affected by spasticity.

As a general conclusion, the present study demonstrates the feasibility of using a robotic device for the evaluation of spasticity in plantarflexors muscles at the ankle joint in children with CP, through the TRST approach. Conversely, the same approach appears to be not consistent for assessing spasticity of the dorsiflexor. Finally, it should be noted that a potential limitation of this approach could be in mapping linear velocity of muscle stretches through angular velocity of the joint. Even though the estimation of the linear velocity of muscle stretches via a neuro-musculoskeletal simulation for assessing spasticity was recently proposed [58], the large amount of scientific researches based on TSRT approach suggests that the use of angular velocity still remains the most widespread methodology for spasticity assessing, due to its easier applicability in clinical routine.

1.4.2. Is there a correlation between the found objective measures and clinical scales?

No correlation between MAS score, TSRT, and μ values was observed. This result confirms the findings reported by Mullick [29] and Jobin [31] about the weak or no-correlation between TSRT and MAS score at the elbow joint, in adults post stroke and children with cerebral palsy, respectively. The absence of correlation might be due to different physiological parameters assessed via the two indices. MAS scale is supposed to evaluate the supra-threshold resistance to passive movement [29], while TSRT measures the limitation in the range of central regulation of the tonic stretch reflex threshold [32], neglecting the supra-threshold resistance [29]. Moreover, MAS depends on the viscoelastic properties of the muscles. Thus, the resistance to external movements could be perceived when no-contractile components of the muscles change [31]. Conversely, variations in the mechanical properties of the muscles are not identified by the TSRT, as they do not cause EMG activity [31]. Considering the different methodological approaches for the evaluation of MAS and TSRT, the absence of correlation between the two indices is not surprising. Consequently, we suggest to use TSRT outputs and MAS score together, enhancing the information about the physiological mechanisms due to deficits in muscle tone.

The observed differences in TSRT values among patients with the same MAS score might be justified by the subject-specific response to the externally imposed joint rotations and the intrinsic inability of the MAS to discriminate neural from non-neural components. The correlation between neural and non-neural components could be evaluated, in future investigations, by examining the biomechanics of the gait. It is generally understood, in fact, that these two components belongs to two different logical levels. Specifically, non-neural components represent a system of constrains and peripheral resources, while neural components utilize these peripheral resources to balance the movement dynamics [59]. Gait analyses might be added to TSRT and MAS evaluations, for a more exhaustive insight in the spasticity phenomenon.

1.5. Conclusions

We demonstrated the feasibility of applying the TSRT approach for the assessment of spasticity of the ankle plantarflexor muscles in children with Cerebral Palsy, by means of a mechatronic device. No correlation was found between the proposed objective measures and the clinical scores based on Modified Ashworth Scale.

Future investigations would integrate the Tonic Stretch Reflex Threshold approach with functional observations, such as gait analysis, permitting a more deeply understanding of the relationship between motor performance and muscles excitability.