Healthcare Sensor System Exploiting Instrumented Crutches for Force Measurement During Assisted Gait of Exoskeleton Users

Matteo Lancini, Mauro Serpelloni, *Member IEEE*, Simone Pasinetti, and Eleonora Guanziroli

Abstract— **Powered exoskeletons can be used by the persons with complete spinal cord injury to achieve bipedal locomotion again. The training required before being able to efficiently operate these orthotics, however, is currently based on the subjective assessments of the patient performance by his therapist, without any quantitative information about the internal loads or assistance level. To solve this issue, a sensor system was developed, combining the traditional gait analysis systems, such as ground reaction force platforms and motion capture systems, with Lofstrand crutches instrumented by the authors. To each crutch three strain-gauge bridges were applied, to measure both axial and shear forces, as well as conditioning circuits with transmission modules and a triaxial accelerometer. An inverse dynamics analysis, on a simplified biomechanical model of the patient wearing the exoskeleton, is proposed by the authors as a tool to assess both the internal forces acting on shoulders, elbow, and neck of the patient, as well as the loads acting on joints. The same analysis was also used to quantify the assistance provided to the patient during walking, in terms of vertical forces applied by the therapist to the exoskeleton. The tests showed a therapist assistance contribution reported as a fraction of the subject body weight up to 40% with an average close to 0% and a standard deviation value of 14%. This paper presents the description of the measurement system, of the post-processing analysis, as well as the results of the proposed approach applied to a single Rewalk user during training.**

Index Terms— **Assistive devices, orthotics, force measurements, rehabilitation robotics.**

I. Introduction

ADVANCEMENT in medical treatment in recent years led to increased survival rates and life span for people with motor-complete Spinal Cord Injury (SCI). However, the quality of life of these patients is still impaired by the mobility limitations and by the decrease in physical health. Moreover, pressure sores, pain, muscle tone, obesity and decreased bone density are all known consequences of the loss of bipedal locomotion (as reported in [1]).

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M. Lancini is with the Department of Mechanical and Industrial Engineering, University of Brescia, Brescia 25123, Italy (e-mail: matteo.lancini@unibs.it).

M. Serpelloni and S. Pasinetti are with the Department of Information Engineering, University of Brescia, Brescia 25123, Italy (e-mail: mauro.serpelloni@unibs.it; simone.pasinetti@unibs.it).

E. Guanziroli is with the Villa Beretta Rehabilitation Center, Valduce Hospital, Costa Masnaga (LC) 23845, Italy (e-mail: eleonora.guanziroli@gmail.com).

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To enable patients suffering from SCI to walk for prolonged time, reducing effort required to patients, different powered exoskeletons were developed, integrating motors on lower limb orthosis, to control hips, ankles, knees and feet (HKAFO) movements. The aim of these wearable robots is to restore a physiological load on the lower limbs, and to reduce the dependence on the arms for propulsion, which could leads to chronic pathologies, as shown by [2]. Since most powered exoskeletons require the usage of walking aids, to initiate or facilitate gait movements, the transfer of part of the weight bearing function to the upper limbs is unavoidable, and should be closely monitored, especially during training.

The training required to learn how to use an exoskeleton for assisted walking is still long and difficult, driven only by subjective evaluations of the performance by the therapist, focusing mainly on the walking rate and the information reported by the patient. The physiotherapist often verifies the correct execution of an exercise by visual inspection only, focusing on a qualitative assessment of movement, drag, synchronism, etc. Moreover, during the training period, the therapist provide assistance to the patient, to facilitate the swing phase initiation and avoid an incorrect swing termination. This is done by following the patient and by manually applying upward vertical forces to the hip section of the exoskeleton, to support part of the patient weight, as shown in Fig. 1 and Fig. 2. Some analytical techniques for quantitative gait analysis could be found in the literature, such as the use of mobile force platforms suggested in [3] or the use of in-shoe pressure monitors reported in [4] and [5]. However, such systems, while sometimes found in gait analysis laboratories, allow measuring loads on the lower limbs of the patient only, neglecting the shoulder involvement, and assess the therapist contribution by visual observation only, without any quantitative analysis.

This paper presents a sensor system that integrates commonly used measurement devices with force sensors developed by the authors [6] and [7]. The sensor system is able to assess internal forces acting on shoulders and elbows thanks to a biomechanical model and a numerical inverse dynamic analysis. The aim of this study is to prove how the approach proposed can be used to measure clinically relevant parameters during assisted gait, focusing on quantities able to improve the user training, such as the upper joints effort and the physiotherapist load bearing role.

As a case study, to test the sensor system, a man suffering from motor-complete SCI, was monitored during his training. The wearable powered exoskeleton used for this case study is a

Fig. 1. A Rewalk™user is supported by his physiotherapist during assisted gait training. The right leg is initiating the swing phase. External forces acting on patient-exoskeleton system are highlighted. F_{TH} and M_{TH} are, respectively the force and torque vectors applied by the therapist. F_{RF} and F_{LF} are the right and left foot ground reaction forces. F_{LC} is the left crutch ground reaction force. F_{RC} the right crutch is not displayed in the picture.

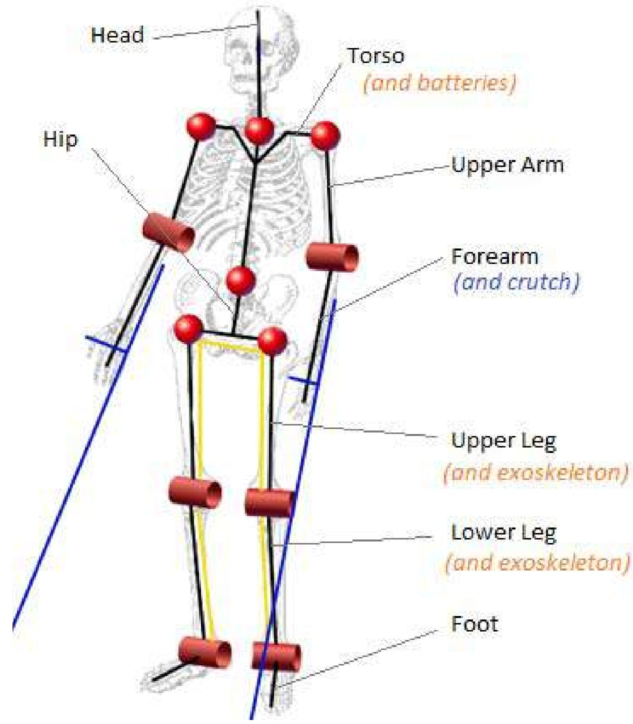

Fig. 2. Mechanical model used to interpret kinematic and dynamic data. Thirteen rigid bodies are used and represented by solid black lines. Red spheres indicate spherical joints, red cylinders indicate cylindrical joints. Crutches are reported in blue solid lines, rigidly linked to the forearms.

Rewalk™unit (Argo Medical Technologies Ltd., Yokneam Ilit, 20692, Israel), that allows over-ground ambulation to subjects with motor complete SCI to walk, but requires the usage of Lofstrand crutches during gait. The robot step movement is initiated by an inclination of the subject torso in the sagittal plane, without any input from the therapist.

II. The Sensor System Proposed

A. Biomechanical Model

As in most cases involving measurements on human subjects, a direct measurement of internal forces acting on upper limb joints is not feasible for exoskeleton users. Therefore external ground reaction forces (GRFs) are used to supply data to a biomechanical model of the exoskeleton-patient system, to have an indirect estimate of internal forces.

The dynamics of the human limbs are complex and not all effects are yet fully understood. Since our application does not have to describe all effects, but to lead to an easily identifiably model should be easily identified we decided to model the patient as a passive multibody system with external torques applied at joints to account for internal forces.

The chosen model reproduces only a simplified version the skeletal portion of the human body, neglecting the muscular components, as well as other soft tissues, in order to minimize the number of parameters to be assessed for each patient, as recommended in [8]. This model was designed starting from a functional model already commonly used in laboratories for kinematic analysis of physiological gait, and adding the minimum modifications needed to describe the assistive devices, to facilitate its adoption by gait lab technicians.

A clear limitation of this simplified model is its inability to capture the complex interactions between the patient and the exoskeleton. While it is true that the SCI patients involved do not control the lower limbs different phenomena are still occurring, such as spasticity, which leads to muscle contractions, as well as velocity-dependent joint stiffness, muscles stiffness, and non-linear damping. Furthermore, the link between the patient and the exoskeleton is not a rigid one: the lower limbs are held in position by fabric straps. Muscles, soft tissues and even garments are interposed between the exoskeleton structure and the patient's bone, allowing for small relative motion between the two systems. The effects of all these phenomena were not taken into account, and cannot be explained by the model.

Thirteen rigid bodies are used to model the human body, replacing each joint group with either a spherical or a cylindrical mechanical joint, as shown in Fig. 2, and considering each crutch rigidly connected to the forearm holding it. A total of 12 joints are used, of which 6 cylindrical and 6 spherical, resulting in 24 rotational degrees of freedom.

Dimensions, mass and inertial properties for each segment were adapted from anthropometric data found in the literature [9]–[11] and scaled to the patient's mass and height, as reported in Table I. Anthropometric data is then integrated

TABLE I
MASS AND DIMENSION OF THE RIGID BODIES OF THE BIOMECHANICAL MODEL

Segment	Length[A]	Mass[B]	Center of mass[C]
Head	18.2%	8.1%	55.0%
Upper arm	18.6%	2.8%	43.6%
Forearm	14.5%	2.2% *+1.5 kg*	68.2% [D]
Upper leg	24.5%	10.0% *+5.0 kg*	43.3%
Lower leg	24.6%	4.7% *+5.0 kg*	43.3%
Foot	15.2%	1.5%	50.0%
Torso	23.8%	34.8%	58.0%
(width)	25.8%	*+3.0 kg*	-
Hip	5.0%	14.9%	14.0%
(width)	19.1%		

These parametrization is based on anthropometric data taken from [10].
[A] As a fraction of the subject height as measured in standing position
[B] Mass is expressed a fraction of the overall subject's mass, plus the fraction of the exoskeleton weight worn by the subject
[C] Proximal distance of the center of mass is expressed as a fraction of the segment length
[D] This value is then corrected by taking into account the mass and center-of-mass of the crutch, considered part of the forearm

by adding the masses of the exoskeleton's segments to the body segments to which they are fixed.

In particular:

1. 5 kg are added to each lower leg to account for the exoskeleton lower leg;
2. 5 kg are added to each upper leg to account for the exoskeleton upper leg and cross beam;
3. 3 kg are added to the torso section to account for the robot batteries, held in the backpack the patient has to wear;
4. 1.5 kg are added to each forearm to account for the crutches mass;
5. The forearm center of mass and inertial properties were computed by simulating the forearm-hand-crutch system numerically with computer assisted design software (Solidworks, marked by Dassault Systemes);
6. Upper and lower leg centers of mass were not corrected.

Given the importance of body segments' parameters on results of the inverse dynamics [12], [13], the segments' length is corrected before any dynamic analysis using kinematic data to assess the distance between joints as suggested by [14].

B. Inverse Dynamic Analysis

After scaling the model to the patient, data from all the three measurement devices is used to perform an inverse dynamic analysis of the patient gait, according to the following steps:

1. An inverse kinematic analysis is used to estimate the values of all the Lagrangian parameters of the model from markers position in time, recorded from the vision system;
2. Impact times of the crutches with ground are compared with the same events reported by the results of the inverse kinematic and the delay between the two measuring systems is corrected;
3. Forces measurements provided by the instrumented crutches are applied to the crutches tip as external constrains for the model;
4. Ground reaction forces, both in magnitude and direction, are added to the model as external inputs applied to the exoskeleton feet;
5. An unknown force and an unknown torque are applied to the hip section of the exoskeleton, with an unknown direction, representing the therapist contribution;
6. An inverse dynamic analysis is performed numerically, computing all forces acting along the constrained degrees of freedom (internal forces) and torques acting along the remaining degrees of freedom (motor torques) as well as the unknown force applied by the therapist (assistance).

These steps were performed numerically using OpenSim (by Stanford University), a specialized multibody modeling software, for simulation and analysis of musculoskeletal models of the human body.

Usually in biomechanical simulations, the unknown force applied to the hip is used to accommodate residual errors between experimental data and simulation [15], and represents the term to be minimized by the numerical solver. In this case, however, forces and torques are actually applied by the therapist to the hip section of the exoskeleton, and with values higher than the uncertainty expected from numerical simulation of the human gait driven by dynamic and kinematic data (a 5 to 95% confidence bound of about 3 N for physiological gait, was reported in [16]).

The closed system, made by the subject, his exoskeleton, and the crutches, exchanges forces and torques only in 6 points, as illustrated in Fig. 1: F_{TH} and M_{TH}, respectively the force and torque vectors applied by the therapist, F_{RF} and F_{LF} the right and left foot GFR, F_{RC} and F_{LC} the right and left crutch ground reaction force.

Taking into account both forces and torques, the model provides 78 dynamic equilibrium equations, 6 for each of the thirteen rigid bodies, and needs to compute 24 unknown motor torques along the degrees of freedom of the model, here reported in Fig. 2. Replacing the model constrains with reaction forces and torques, other unknowns are added to the problem: 18 internal forces for the 6 spherical joints, 30 internal forces for the 6 cylindrical joints, totaling 72 unknowns.

To compute these unknowns, as well as F_{TH} and M_{TH} (with three parameters each, totaling 78 unknowns), the numerical solver needs to be provided with the values of all ground reaction forces (F_{RF}, F_{LF}, F_{RC} and F_{LC}) which are obtained by direct measurement, along with kinematic data.

C. Gait Lab Measurement Equipment

Kinematic measurements are performed using a marker-based Smart-dx vision system (by BTS) made of 8 wall-mounted infrared video cameras, with a sampling frequency of 2 kHz, a resolution of 4 Mpixels, and a position accuracy, as declared by the manufacturer of 0.1 mm, for the whole walking range of 6 m.

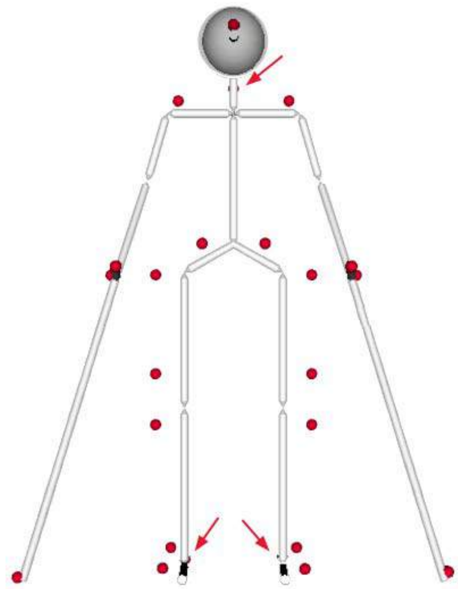

Fig. 3. Position of the passive markers (red dots) on the subject. The red arrows indicate marker on the back of the patient (red dots not visible).

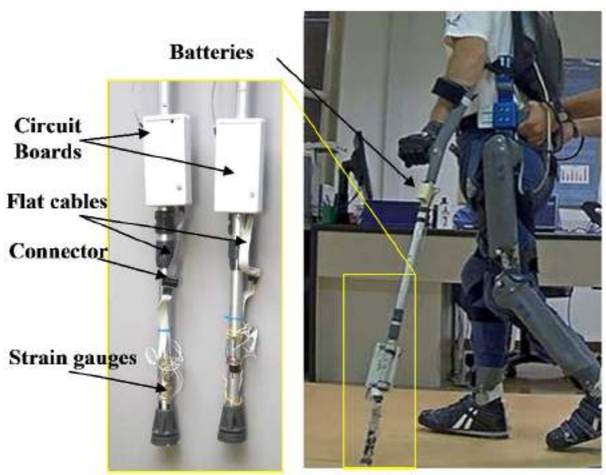

Fig. 4. Detail view of the lower part of the instrumented crutches used. Circuit boards, strain gauges and batteries are indicated.

Twenty-four retro-reflective spherical markers with a 20 mm diameter are applied on both the patient and the exoskeleton following a modified Davis protocol [17]. To get a complete assessment of each degree of freedom of the adopted model, three markers are positioned on each crutch, 7 on the patient and 12 on the exoskeleton, as reported in Fig. 3.

Ground reaction forces (GRF) between the patient's feet and the laboratory ground are measured using eight P-6000 load platforms (by BTS), split in two lines of four each, and hidden in the floor. Each platform line can host one foot and the proprietary software used to collect their data records the resulting GRF vector, for each platform line, with a 6 N accuracy on each axis (approx. 0.6% of the patient body weight).

D. Instrumented Crutches

A pair of instrumented Lofstrand crutches capable of measuring in real time axial forces, shear forces, angles of inclination and time of impact on the ground have been used in the tests (Fig. 4). Twelve strain gauges forming three Wheatstone bridges in full-bridge configuration are mounted on the mechanical structure of the crutch. The three bridges allow measuring axial and shear forces.

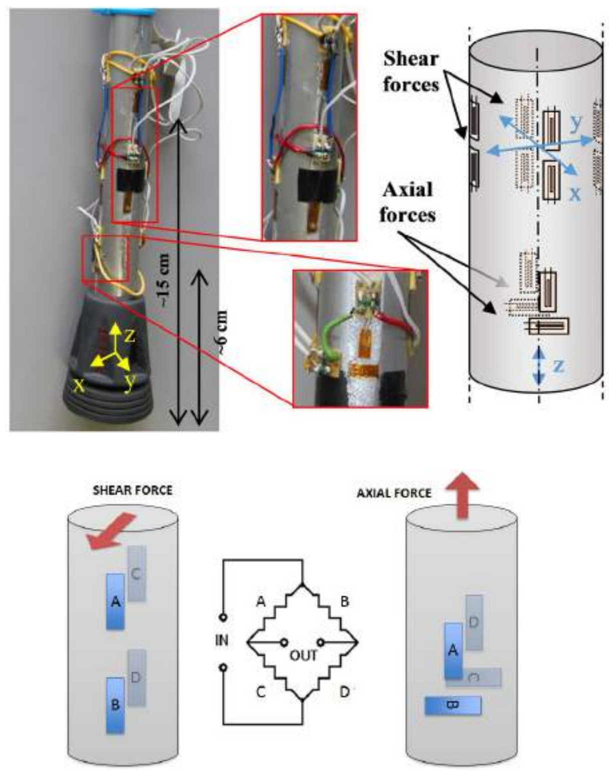

Fig. 5. Position of the strain gauges on the crutchs (top left), reference axes (top right) and bridge connections scheme (bottom).

For the three bridges, strain gauges marketed by RS, with a resistance of 120 Ω and an active length of 2 mm, were used. The strain gauges were mounted on the bottom of the crutch with a based ethyl-cyanoacrylate adhesive (from Omega) and then covered with a protective mastic silicone (from HBS) to preserve the integrity from accidental impacts.

The strain gauges are applied according to the diagram shown in Fig. 5. The axes orientation follows the conventions of the International Society of Biomechanics (ISB), with Z along the main axis of the crutch, X along the handle and Y accordingly, thus creating the classic reference system [18]. The axial forces are measured by using the full bridge configuration with two strain gauges along Z-axis of the crutch, while two others along Y, compensating the apparent deformation due to temperature changes and shear forces. While the shear forces are measured using two different full bridges, one normal to X and the other in the surface Y, normal to X. For both of these bridges, the strain gauges are aligned to the Z direction and the distance between them is about 10 mm. This configuration allows compensating temperature effects.

On each crutch is also mounted a circuit board, which includes a conditioning circuit with wireless transmission module, a triaxial accelerometer (LIS3LV02DL - STMicro). used as tilt sensor, and a circuit for power management. The whole system is powered by rechargeable batteries. Each

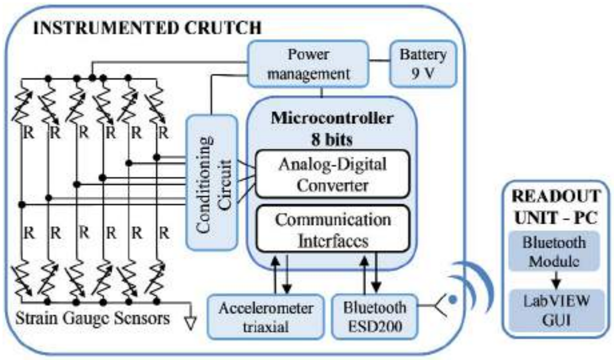

Fig. 6. Block diagram of the circuit board for each instrumented crutch.

printed circuit board is attached to each crutch using a plastic removable box to perform maintenance easily. The circuit board is connected to the strain gauges through a detachable flat cable (Fig. 4).

A block diagram of the circuit board is shown in Fig. 6. This configuration is similar to the system reported in [6]. An eight-bit microcontroller (S08AC128-Freescale) manages conversion and data communication with the readout unit. The strain gauges are connected to a conditioning circuit to filter and amplify the sensors signals, which are then acquired using the integrated 10-bit analog-to-digital converter (ADC). Furthermore, the microcontroller controls the triaxial accelerometer via a serial peripheral interface (SPI), whereas the wireless data communication to the host computer is obtained by the microcontroller sending the data via a serial communication interface (SCI) to a Bluetooth module (Parani ESD200) soldered on the circuit board. The readout unit consists simply of a PC (personal computer) with a commercial Bluetooth module. The communication on the PC is managed by a specifically developed Virtual Instrument (VI) using LabVIEW™.

The circuit board is powered by two 9V batteries mounted on each crutch and connected in series. The battery voltage is regulated by DC-DC regulators to obtain a fixed voltage of 5 V for the conditioning circuit and 3.3 V to power supply Bluetooth and accelerometer.

The microcontroller's firmware is written in C using CodeWarrior software from Freescale, and it is identical for both crutches. At first, the firmware loaded into the microcontroller executes initialization of variables and peripherals. Then, after the communication between crutches and readout unit is established, the firmware reads every 30 ms the data from the accelerometer, acquires the signals from the three bridges and sends all the data to the Bluetooth for transmission to the PC.

The measurement data from both crutches are collected using a PC with a commercial Bluetooth USB dongle and a virtual instrument developed in LabVIEW (by National Instruments). Measurement data from both crutches are collected, with a 30 ms sampling interval, by the virtual instrument, which records forces in a spreadsheet format for post-processing, as well as displaying the values in real time, to allow for a direct feedback during tests.

Tilt angles are also recorded, only to identify when the crutches touch the ground. Since there is no way to use a hardware or software trigger to synchronize the force measurements obtained by the instrumented crutches with the other elements of the gait measurement framework, this information is used to correct, in post processing, any delay between the crutches recording and the rest of the measurement system. The impact of each crutch with the ground is, in fact, an event clearly identifiable in all the system involved: in crutches acceleration measurements is seen as an acceleration peak, in crutches tilt angle measurement as a sudden variation of angles at the end of a smooth sinusoidal motion, in crutches forces as a sudden peak from zero, and in kinematic measurements performed using the motion capture system, as both the minimum vertical position of the lower crutch marker and as the end of a rotation of the crutch-forearm system around its Y axis.

Each crutch was characterized in the laboratory. A mechanical system built ad-hoc was used for the characterization. Each crutch was set along each axis and then repeatedly loaded with known masses. For compression tests, static compression were applied, with different masses up to 60 kg ($\sim$600 N). For shear tests, each crutch was connected horizontally to the mechanical system, and then static loads were applied to the crutch tip, using different masses up to 15 kg ($\sim$150 N) in both directions. While the characterization of angular measurements was done via an angular micrometer screw.

The orientation of each crutch, both in the sagittal and frontal plane, is measured by a triaxial accelerometer Typically, Rewalk™users walk with an average speed of 0.5 steps per second, which is almost half the physiological value, therefore it can be assumed that the movements of the crutches are performed with a low dynamic range. This allows, according to [19], to ignore the dynamic components of acceleration. It is therefore possible to calculate angles from the projections of the gravity vector on each accelerometer axis. This hypothesis is always valid except when the crutch tip touches the ground, when impact accelerations may (in some cases) not be negligible. Since this behavior easily allows the automatic identification of the impact time of each crutch, using a simple threshold, this solution has been implemented.

In Fig. 7, the results of characterization tests performed on the two crutches are shown. The measurement data show a non-linearity of 40 mV, and three standard deviations (3SD) of about $\pm$0.1 V for all three axes and $\pm 0.3°$ for angle measurements.

III. Results

Internal forces, motor torques and therapist assistance are all computed at the same time by the inverse dynamic analysis, but the main objective of this study is to demonstrate the possibility to measure two key quantities: the internal forces acting on shoulders and the therapist assistance. The former is needed to avoid overloading the patient upper limbs, the latter to provide information to the therapist about the level of assistance provided during training.

Being unable to perform reference measurements of the same quantities (shoulders loads and therapist assistance),

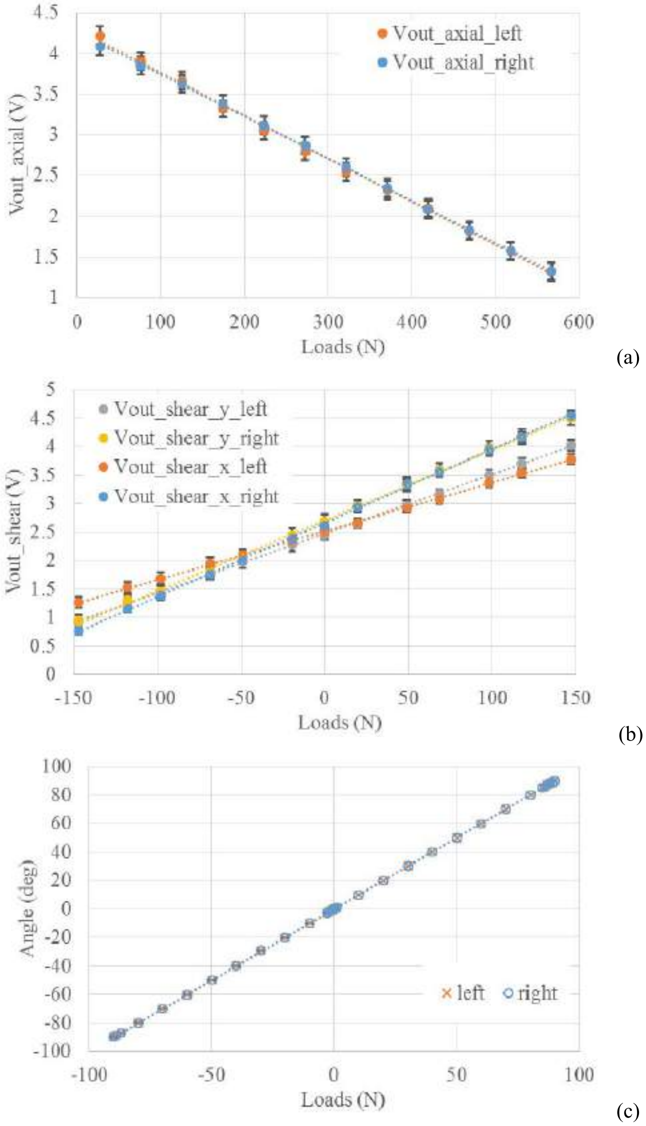

Fig. 7. Test results for a sample crutch for compression loads (a), shear forces (b) with known masses applied and (c) tilt angles.

without interfering with the patient's training, a preliminary evaluation of the measurement system performance was limited to the repeatability of the phenomena measured during different walks. To limit variability due to difference between patients (level of lesion, age, weight, sex...) and due to the small number of SCI patients training with the same exoskeleton, a single subject was involved in the study.

The proposed approach has been applied to a Rewalk™user suffering from motor-complete SCI during the last part of his training in assisted walking with the exoskeleton. The patient was asked to walk at a comfortable speed along the force platform lines, while the motion capture system recorded his movement, using the instrumented crutches presented as a walking aid. The patient was closely followed by a physiotherapist, instructed not to step onto the force platforms himself, to avoid any incorrect GRF measurement. The procedure has been repeated ten times.

TABLE II

INTERNAL FORCES ACTING ON UPPER JOINTS AND THERAPIST ASSISTANCE

Joint	Maximum	Average	RMS*	SD[A]
Right shoulder	30%	10%	13%	7%
Left shoulder	28%	9%	11%	7%
Right elbow	33%	11%	14%	8%
Left elbow	30%	10%	12%	7%
Neck	8%	3%	3%	1%
Hip-Torso	57%	17%	20%	11%
Therapist assistance	*41%*	*1%*	*14%*	*14%*

Results from measurements of 10 step in 5 different walks of a Rewalk™ user under the same environmental conditions. Only the two central steps for each walk were taken into account. Results are normalized by dividing for the subject body weight.

* Root mean square value of the samples of the two central steps

[A] Standard deviation of the samples recorded during the two central steps.

The two central steps of five walks (not including the first and last one) were analyzed. Following clinical conventions, those forces were normalized dividing their values by the patient's weight. Synthetic results obtained from these 10 steps are reported in Table II. The maximum value is about 30% of the body weight for shoulders and elbows, while the Torso-Hip joint accommodates nearly 60% of the body weight. Average values are about 10% for upper joints, showing how their overall involvement in this patient usage of the exoskeleton is correctly limited. Since the process is dynamic, the root mean square value (RMS) of the forces involved was also computed, with values about 10% for upper joints, a negligible value (3%) for the neck, and 20% of the body weight for the hip-torso joint.

The therapist assistance, measured as the vertical upward force applied to the exoskeleton by the therapist, is also reported as a fraction of the subject body weight, and shows a contribution up to 40%, but with an average close to 0% and a RMS value of 14%.

To represent the evolution in time of the measured quantities, forces computed using the proposed approach from five walking tests were normalized using as a reference the right heel contact time, given 100% the time between a right heel ground contact to the next one. The two central steps of the five walking tests were then linearly interpolated. For each 0.1% stride cycle resolution the average and standard deviation of forces were computed, and they are here reported in Fig. 9 and Fig. 10.

Average heel contact and toe-off events, as detected by the inverse kinematic analysis, are also reported, to facilitate interpreting the results. Left heel contact and left toe-off events times registered a 3% standard deviation, the first right toe-off showed a 3% standard deviation, while for the second right toe-off a 5% standard deviation was shown. Such variability was not represented to make the charts easier to read, but should be taken into account.

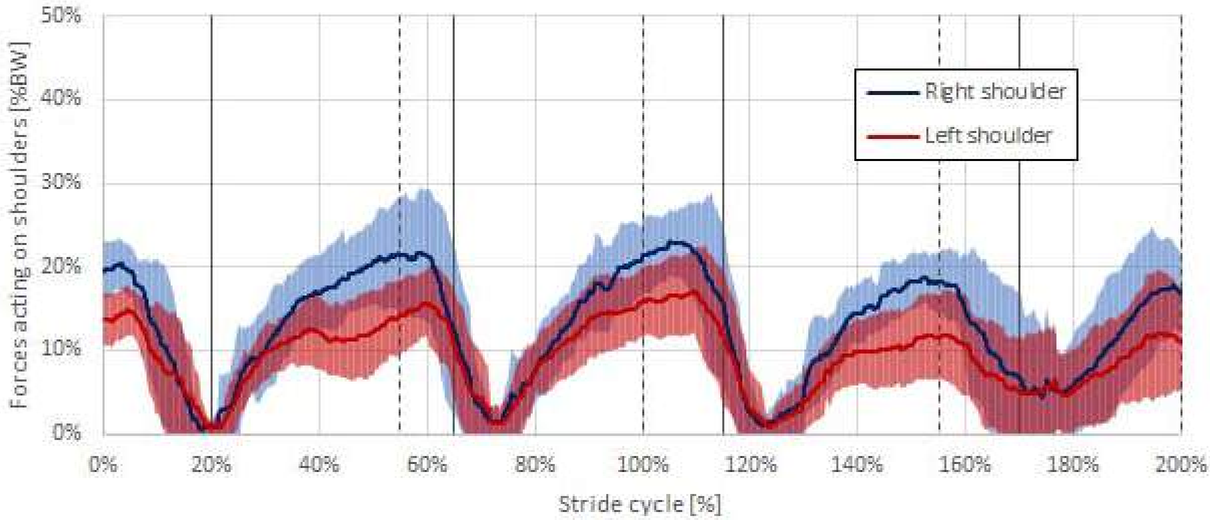

Fig. 8. Loads acting on shoulders as fraction of the body weight, during two steps (given 100% the time between a right heel ground contact to the next one) for five different walks of the same patient in the same environmental conditions. The red line indicates the average left side internal forces, the blue line is for the right side, the background of the same color indicates a confidence interval of one standard deviation. Vertical dotted lines indicate heel to ground contact, vertical solid lines indicate toe detachment from ground. The period between a vertical solid line and a vertical dotted line indicates a swing phase for a lower limb (e.g. between 20% and 55% the left leg is swinging, between 65% and 100% the right leg is).

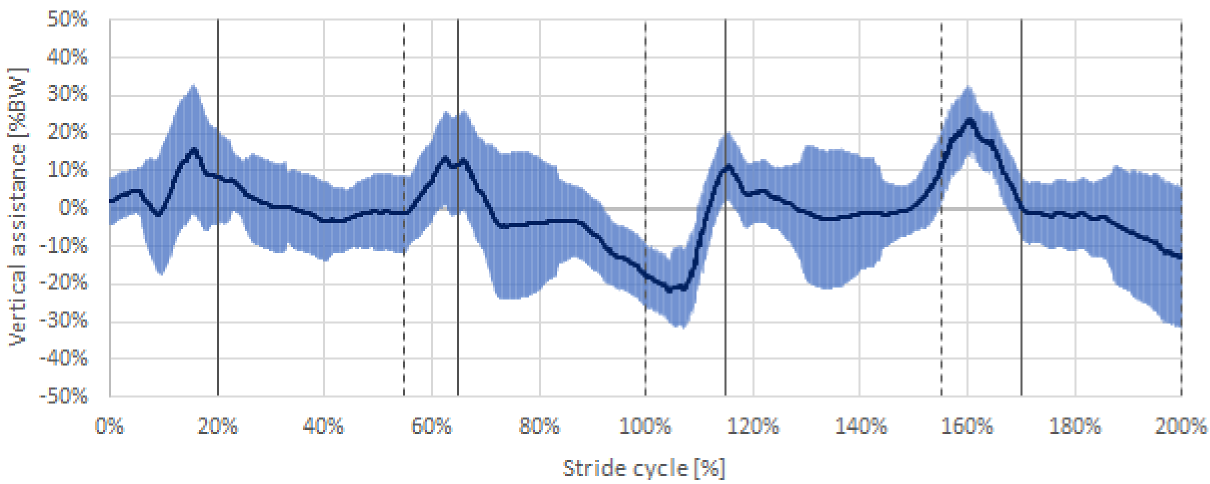

Fig. 9. Physiotherapist assistance level as fraction of the body weight, during two steps for five different walks of the same patient in the same environmental conditions. The blue solid line indicates the average vertical upward force applied by the therapist to the exoskeleton hip section, its blue background a confidence interval of one standard deviatiion. Vertical dotted lines indicate heel to ground contact, vertical solid lines indicate toe detachment from ground. The period between a vertical solid line and a vertical dotted line indicates a swing phase for a lower limb (eg., between 20% and 55% the left leg is swinging, between 65% and 100% the right leg is).

In Fig. 8 the forces acting on the upper limbs of the patient are visible, showing how an average of 20% of the body weight is sustained by each shoulder at the end of each swing phase, while in most cases this value does not exceed 15%, also highlighting differences between the left and the right swing phase in terms of loads. The standard deviation recorded has an average value of 5% (of the body weight) for the right shoulder, and 6% for the left one, but reaches 10% for both sides during the double support phase (after one heel contact and before the contralateral toe-off).

Fig. 9 shows the vertical upward force applied by the therapist to support the patient during his walk. A periodic assistance with average peaks of 20% of the body weight is visible, with higher values recorded before each swing initiation, probably to facilitate a correct detachment of the foot from the ground in the absence of a lateral movement by the patient himself.

At the end of both steps in all walks, a negative assistance is visible, likely because the therapist accidentally pushed down the exoskeleton to avoid falling after a temporary loss of his balance due to a sudden movement by the patient.

The therapist confirmed this, explaining how this often happens just before lifting the patient to help his left leg swing initiation, due to the fact that the therapist is forced to walk

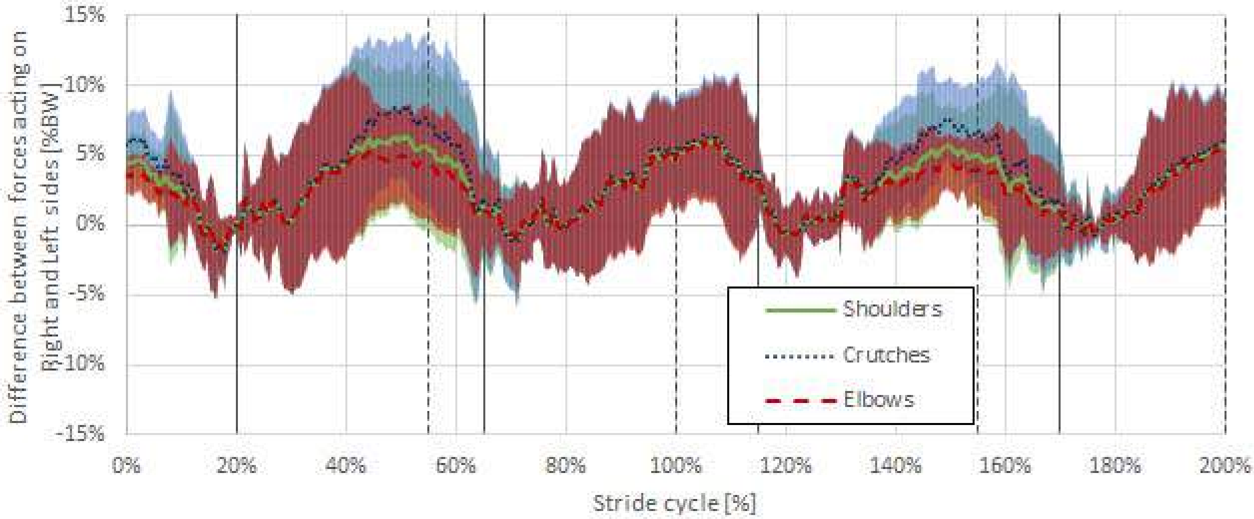

Fig. 10. Difference between the right side and left side forces as fraction of the body weight, during two steps for five different walks of the same subject in the same enviornmental condition. The dotted blue line indicates the average difference between forces acting on the right and left crutches, the solid green line is for the average difference of forces acting on right and left shoulders, while the dashed red line is for the elbows. The backgroud in the same color indicates a confidence level of one standard deviation. Vertical dotted lines indicate heel to ground contact, vertical solid lines indicate toe detachment from ground. The period between a vertical solid line and a vertical dotted line indicates a swing phase for a lower limb.

with spread legs to avoid the force platforms positioned on the two central steps, and rarely occurs when force platforms are not used.

The standard deviation recorded has an average value of 10% (of the body weight), but reaches 20% around the toe-off event of both feet. To assess the impact of the model parameters describing mass distribution, a second simulation was performed with the same experimental data, but with a mechanical model with mass centers intentionally moved 50 mm from their original position (along the longitudinal or lateral direction, away from the patient). The maximum difference between the assistance level computed by the two simulations amounted to 3% of the body weight, far less than the 10% variability (standard deviation) between different steps of the same subject.

Another analysis made possible by the measurement of internal forces, is the investigation of the difference between forces acting on the right and left side. Since an asymmetric load of upper limbs joints could lead to back pain and other dysfunctions, as reported in [21], it is important to monitor this parameter, especially while the patient learns how to properly use his wearable robot.

A detail of this difference, computed for the loads recorded directly by the crutches and the internal forces acting on elbows and shoulders for five different walks of the same subject, is visible in Fig. 10. As can be noticed, the values are generally below 10%, and a prevalence of the right side is visible. The subject was right-arm dominant.

It is interesting to notice how the values directly reported by the crutches, for what concerns the load differences, are close to the same values for the upper joints internal forces, suggesting the idea that the usage of the crutches alone could provide an approximate assessment of the shoulders symmetry.

The Symmetry Index proposed in [20], was also computed following equation (1), where FD and FnD are the mean values

TABLE III

DIFFERENCE BETWEEN FORCES ACTING ON RIGHT AND LEFT SIDE JOINTS

Joint	Min	Max	Average	SD[A]	**Si[B]**
Shoulders	-14%	16%	1%	6%	**12%**
Elbows	-15%	18%	2%	7%	**11%**
Hips	-16%	22%	0%	11%	**2%**

Results from measurements of 10 steps in 5 different walks of a Rewalk™ user under the same environmental conditions. Only the two central steps for each walk were taken into account. Values are normalized by dividing for the subject body weight. A positive value indicates a right side prevalence.
[A] Standard deviation of the samples during the two central steps.
[B] Symmetry index computed using the average value of 10 peak forces for the dominant and non-dominant upper limbs.

of the peak forces recorded for the dominant and non-dominant limb, respectively. As suggested by the referred authors, $-10\%<\mathrm{SI}<10\%$ indicates symmetry, while $\mathrm{SI}>10\%$ or $\mathrm{SI}<$-10% indicates asymmetry.

$$Si = \frac{FD - FnD}{FD + FnD} \times 200. \tag{1}$$

Both the symmetry index and difference between right and left forces were computed for ten steps in 5 walks of the same subject, and they are synthetically reported in Table III. As can be noticed, the average values of force difference display a very limited prevalence of the right side (below 3%) and show higher differences only when comparing the minimum and maximum values, while the symmetry index indicates an asymmetry in the loads acting on the upper limb joints.

IV. Conclusions

Kinematic analysis, focusing on gait rate and swing/stance timing, is commonly used in gait laboratories, and is currently the main measurement, with electromyography signals analysis, used to monitor patients using powered exoskeleton as locomotion tools. On the other hand, forces and torques are only marginally used due to the complexity of their measurement in a clinical setting.

Motor torques have a limited interest during assisted gait, because the patient directly controls them and should be able to avoid excessive efforts. In contrast, internal forces, given by constrain reactions in the joints, should be closely monitored to avoid overloading the joint groups, such as the shoulder, both during training and in everyday use.

This sensor system could be used to identify any anomaly in the patient gait, such asymmetry in the load bearing functions, or asynchrony between different sides. This will help physiotherapist to plan specific correction exercises and guide the patient during training.

However, measuring torques and forces acting in human joints is a difficult task, which requires indirect measurements and specific mechanical models to transfer direct measurements of interaction between the human body and the ground, to the actual measurand of clinical relevance. All forces exchanged between the body and the environment have to be taken into account to obtain a reliable measurement.

The high variability between different walks of the same subject should be investigated as a further development of the study. In particular the anthropometric parameters used could be estimated using numerical optimization based on the difference between simulated results and a directly measurement of the therapist assistance. This approach, proposed by the authors but not currently employed due to hospital insurance limitations, could also be used to provide a proper metrological validation of the system.

In the presented case study, a powered exoskeleton user supports himself with forearm crutches, this means adding to the model two elements interacting with the ground, to account for the crutches, and an external force at the hip, to account for the therapist presence and assistance.

Used alone, the instrumented crutches developed can help avoiding excessive or asymmetric loading of the upper limbs, for Rewalk™users, even in a foreseeable domestic setting. This could help to reduce or to avoid the long-term effects of assisted locomotion with walking aids, and to prevent the subject from misusing his wearable robot.

Moreover, if the presented crutches are integrated with traditional gait lab equipment, as shown in this paper, they can be used to provide information on internal forces and torque during assisted gait, and become a complete sensor system, currently not yet available in gait labs, able to quantify the physiotherapist level of assistance.

This quantitative measurement can offer an insight into the training progress, helping physiotherapists to assess both the patient performance, and their contribution to it. This will allow to adapt and to correct patient-therapist interactions, leading to a faster and safer training for exoskeleton users.

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Matteo Lancini received the master's degree in mechanical engineering from the University of Brescia, Brescia, Italy, in 2005.

He received the Ph.D. degree in applied mechanics from the University of Brescia in 2015. He was a Research Contractor with the Department of Industrial and Mechanical Engineering, University of Brescia, where he was involved in the development of measurement techniques for industrial diagnostics, and focused on measurements in uncontrolled environments from 2005 to 2009.

He is currently an Assistant Professor with the Mechanical and Thermal Measurements Laboratory, Department of Industrial and Mechanical Engineering, University of Brescia. His current research interests include measurement systems for biomechanical analysis, in particular, for robotic gait and rehabilitation, and industrial diagnostics using nondestructive techniques based on vibration measurements.

Dr. Lancini is a member of the International Society of Biomechanics.

Mauro Serpelloni received the Laurea (*summa cum laude*) degree in engineering and the Research Doctorate degree in electronic instrumentation from the University of Brescia, in 2003 and 2007, respectively. He is currently an Assistant Professor of Electrical and Electronic Measurements with the Department of Information Engineering, University of Brescia. He was involved in several projects relating to the design, modeling, and fabrication of measurement systems for industrial applications. His research interests include biomechatronic systems, contactless transmissions between sensors and electronics, contactless activation for resonant sensors, and signal processing for microelectromechanical systems.

Simone Pasinetti received the B.S. and M.S. (Hons.) degrees in automation engineering from the University of Brescia, Brescia, Italy, in 2009 and 2011 respectively, with a thesis concerning the control of mechanical actuators with SEMG signals, and the Ph.D. degree in applied mechanics from the University of Brescia, in 2015, with a thesis titled "Development of Measurement Protocols for the Analysis of the Functional Evaluation and Rehabilitation, in Biomechanics Field." While working towards the Ph.D. degree, he was with the Institute of Intelligent Systems and Robotics, Paris, France, where he carried out research concerning the dynamic posture analysis.

He has been a Research Fellow with the Optoelectronics Laboratory, Department of Information Engineering, University of Brescia, since 2015. He is currently involved in posture analysis and liquid lenses system development. His research interests include biomechanics of human posture, EMG analysis, and 2-D and 3-D vision system development.

Eleonora Guanziroli received the Ph.D. degree in applied mechanics in 2014 and the master's degree in biomedical engineering in 2008. She is currently a Consultant Researcher with the Bioengineering Division, Villa Beretta Rehabilitation Center, Valduce Hospital. She is with the Movement Analysis Laboratory and follows research activity in biomedical fields. The research activity focuses on the development of robotic devices aimed at walking recovery of neurological patients, application of robotic devices in rehabilitation fields, detection of physiological parameters on neurological patients during robotic training, application of functional electrical stimulation during a reaching movement assisted by robot, application of functional electrical stimulation for the correction of drop foot, software development in MATLAB environment for the integration of different signals (electromyography, kinematic data, and so on), description of sitting posture using quantitative and qualitative data, and management and maintenance of the electronic medical record. Her research activity also includes European Projects and International Cooperation Projects. She is a Study Coordinator with the Multicenter Projects about the use of botulinum toxin where knowledge of ICH-GCP and awareness of local regulatory regulations regarding drug research is required.