2021 43rd Annual International Conference of the
IEEE Engineering in Medicine & Biology Society (EMBC)
Oct 31 - Nov 4, 2021. Virtual Conference

A Sensorized Overground Body Weight Support System for Assessing Gait Parameters During Walking Rehabilitation

Bruno Leme*[1], Chun Kwang Tan[1], Eleuda Nunez[1],
Masakazu Hirokawa[1], Kenji Suzuki[1,2], Hideki Kadone[2]

***Abstract*— Although the needs of individuals undertaking gait rehabilitation sessions may appear similar, they present facets that may assist therapists to come up with more targeted treatment. However, acquiring such aspects is a major problem for rehabilitation personnel due to time constraints and/or complexity. In this paper, we propose an alternative method for estimating gait parameters for individuals requiring Body Weight Support (BWS) during gait training. Results show that the proposed device is able to acquire step length and the amount of body weight unloaded with relatively high accuracy. This reduces the need to set up external sensors to measure patients. Moreover, it can provide gait parameters for patients evaluation which can be used for more personalized treatment.**

***Clinical relevance* - Tracking patient progress during therapy is an important part of personalized therapy. The proposed device is a simple, low-cost method of collecting gait parameters from patients, without the use of expensive motion tracking and force sensors.**

I. INTRODUCTION

For individuals with impaired gait, such as stroke and traumatic brain injury survivors, walking rehabilitation is a necessary part of recovering their functional capabilities. This involves the use of assistive devices and the aid of therapists to stimulate them to relearn motor skills and to recompose muscular tenacity [1].

Recently, a growing number of papers have claimed the benefits of personalized medicine to boost the recovery of patients in physical rehabilitation [2], [3], [4], [5]. The idea is to have a deeper understanding of each patient's condition, such as their limitations and progress, to propose the most appropriate treatment. There are many qualitative methods used to evaluate gait quality (e.g. Fugl-Meyer Assessment (FMA), Functional Independence Measure (FIM), Manual Muscle Test (MMT)). Besides, some medical devices can be used to obtain quantitative information. Motion capture system, for instance, can be used to evaluate many kinematic parameters of gait [6]. Also, force plates can be used to assess kinetic gait parameters such as balance and gait symmetry [7]. Moreover, the combination of other sensory modalities to evaluate gait can be applied to provide a better diagnose. Ramakrishnan et al., for instance, highlights kinetic anomalies on the gait of walking impaired individuals when walking with symmetric step length [8].

* Corresponding Author: leme@ai.iit.tsukuba.ac.jp
[1] University of Tsukuba, Faculty of Engineering, Information and Systems
[2] Center for Cybernics Research, Faculty of Medicine, University of Tsukuba

Acquiring gait data during walking rehabilitation presents many challenges. Clinical evaluation tests are typically performed as part of a patients' therapy session, and occupies its own time segment in the therapy session. Moreover, some of the evaluations are highly subjective and based on the experience level of the therapist, which may lead to conflicting diagnosis by a different therapist. On the other hand, laboratory-based evaluation methods, like motion capture and ground reaction force (GRF) analysis, requires time and special equipment to set up, which is not what a typical therapist have access to. Furthermore, analyzing the data requires a different set of expertise which a typical therapist is not trained in. Hence, such methods might not be practical in a general sense.

In this paper we propose a quantitative method of evaluating the progress of patients during walking rehabilitation. Furthermore, considering the personnel limitation during rehabilitation and the burden for patients for placing external sensors, we propose a ready-to-go device which can measure step length and amount of body weight unloaded information without much posterior analysis. The amount of body weight unloaded will be used to calculate GRF and a gait symmetry index based on GRF.

II. METHODS AND MATERIALS

A. Estimating step length

During walking rehabilitation, some gait parameters can be used to estimate the progress of patients. A Motion Capture System (MoCAP), for instance, is commonly used to assess body kinematics of individuals, which offers a vast range of information with high precision. On the other hand, the preparation of markers and the post-processing necessary is mostly impractical due to limited personnel. Furthermore, for patients who depend on a walking harness to support part of their body weight, the placement of markers represents an extra challenge due to the belts covering the pelvis and occlusions caused by the device.

The number of steps and the distance walked can be an indicator of the progress of the treatment. Our previous work [9] uses a Laser Range-Finder (LRF) to detect the feet position during walking. To avoid losing track of participants' feet during gait transition, the sensor was placed 40cm above the ground. On the other hand, the selected position results in a fractioned reading of the real step length requiring further calculation. Since our aim is the maximum step length during gait cycle, which is during Double Limb Support (DLS), we disconsider the knee flexion and model the human gait such as the inverted pendulum model [10]. Figure 1 illustrates the model considered and the positioning of the LRF.

To estimate the real step length, the hip joint angle (hip flexion and extension angles) (θ) on the sagittal plane, which is the angle formed by the two legs, needs to be calculated. It can be found by interactions using the following equation:

$$\frac{D}{2 * sin(\theta)} + \frac{H}{cos\theta} - L = 0 \tag{1}$$

Where θ is the hip joint angle, D is the distance measured between the two legs using the LRF, H is the sensor high from the ground, and L is the leg length. Finally, the real step length (SL) can be estimated using the hip joint angle and leg length by using the following equation:

$$SL = 2 * L * sin(\theta) \tag{2}$$

B. Estimating Ground Reaction Forces

The ground reaction force can be used to evaluate several gait parameters. The leg to leg vertical load transition, for instance, can be used as a metric to evaluate symmetry. Moreover, it may be important to assess the real amount of body weight support provided, which could allow more precise control of patients' unloading. Finally, the load supported by patients during walking over time can indicate their muscular strength and consequently their progress among sessions.

As an alternative to force plates, strain-gauges were used in our platform to measure vertical and horizontal strains. It was installed perpendicular to the cross section of the harness' supporting fork (longitudinal to the tube). To provide temperature compensation, extra strain gauges were placed nearby our initial sensors with 90deg phase (transversal to the tube). A half-bridge Wheatstone circuit was implemented near the strain gauges and HX-711 transducers were used to convert the deformation. A microcontroller was used to acquire the information from the transducers using I2C protocol and communicates with a computer by using TCP/IP. After confirming the linear behavior of the strain-gauges, a module was developed for calibration.

The circuit bridge was implemented allowing that vertical descendent forces receive positive readings on each side of the fork. When supported by the harness, both sides of the

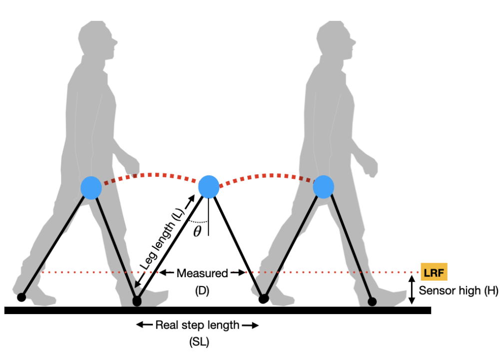

Fig. 1. Proposed model and position of the LRF to estimate the real step length

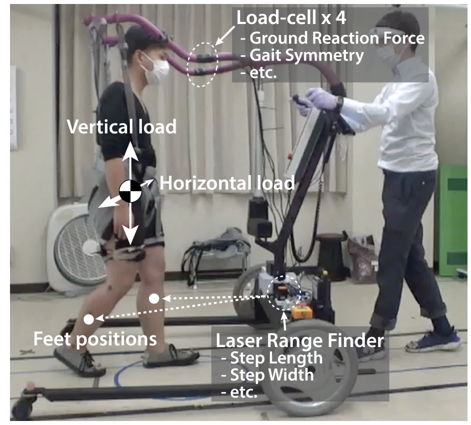

Fig. 2. Overview of the proposed device.

fork are pulled due to the angle formed by the belts and the weight of the user. The direction of the horizontal forces was defined so that the sum of these readings tends to cancel one another.

A software was developed in c++ to receive the readings and convert them into horizontal and vertical loads. To calculate the resulting horizontal load the following equation is used:

$$H_{load} = (H_{left} + H_{right}) + offset \tag{3}$$

where H_{load} is the horizontal resulting force, H_{left} and H_{right} are the left and right horizontal loads read by the strain gauges and the offset is the initial calibration of participants. Considering the patients who need to rely on the harness for body weight support while walking, the vertical ground reaction force (GRF), can be obtained using the following equation:

$$GRF = W_{user} - (V_{left} + V_{right}) \tag{4}$$

Where W_{user} is the user's total body weight, and V_{left} and V_{right} are the vertical left and right loads read by the strain gauges. The total amount of GRF is obtained as an absolute number with no correlation with feet condition. Considering that during human gait the CoM is constantly transferred from leg to leg, the period where it is divided in both legs during DLS phase is neglected. Therefore, the gait peaks and valleys acquired from the LRF are used to indicate the moment in which the GRF has to be assumed as the right and left side respectively.

The overall platform with modifications can be seen in Figure 2. It has an on-board computer for processing the data and to display feedback to the therapist. A server was implemented to record the different data into a single timestamp and its access is easily possible in a remote server using WiFi.

C. Evaluating symmetry

By having GRF and step length simultaneously during walking, some extra evaluations can be proposed. Kinetic symmetry, for instance, is proposed by some authors using GRF obtained by force plates [11], [7]. We use the same method, in combination with the proposed sensors, to estimate the symmetry of participants during all the sessions. The Symmetry Index (SI) can be found by using the following equation:

$$SI = 100 * \frac{G_{left} - G_{right}}{0.5 * (G_{left} + G_{right})} \tag{5}$$

where G_{left} and G_{right} are the GRF on the left and right side respectively.

III. Experimental Protocol

Three experiments were conducted to evaluate our proposed device. Two experiments were intended to compare the designed sensors with gold-standard measurement methods. The last experiment was conducted to estimate gait parameters using the new sensors. The experimental procedures involving human subjects were approved by the Institutional Review Board of University of Tsukuba Hospital.

A. Step length

Measurements acquired by our platform was confronted with a motion capture system (VICON MX System with 16 T20S Cameras, 100hz capture, Vicon, Oxford, UK). Markers were placed in the left and right heels of two healthy participants. Furthermore, visual indications were placed on the floor composed of 3 lines distanced 50cm from each other and 3 lines distanced 25cm from each other. Participants were instructed to step according to the visual marks reproducing the step length described and after continue walking at free step length and cadence. The processed step length data was recorded and later compared to the motion data to estimate the error.

B. Body loads

To evaluate whether our proposed strategy for acquiring body loads is consistent with the real forces involved. Two force-plates (Model: ACG, AMTI, MA, USA) recording at 1kHz were used while two healthy participants were supported by the harness in different conditions (25%, 50%, and 75% of BWS). Due to the number of force plates available and also the difficulty imposed by harnesses to measure body weight during walking, the force plates were placed side by side and participants were instructed to simulate walking. The vertical and horizontal loads on an average of ten steps were recorded by both systems simultaneously. The peaks of each step was obtained and the mean, STD and RMSE was calculated.

C. Gait symmetry index

The gait symmetry and amount of body weight unloading of four healthy participants were evaluated. To simulate asymmetric gait, a patellar-tendon bearing orthosis (PTB) was used to constrain the angle of motion and reduce the pressure on the foot sole. The experiment consisted of participants walking with different unloading (25%, 50%, and 75% of BWS) without leg constrain and after in the same condition using the orthosis. The decision on which leg to constrain was randomly selected among participants. They were instructed to walk at their desired step length and speed and to rely as little as possible on the device to walk.

IV. Results

The peaks in the step lengths were acquired from the LRF and the motion capture data in each condition. The mean and STD were calculated for each walking condition. Also, the RMSE was calculated comparing the data of the two sensors. The results were compiled in table I.

TABLE I
Step length comparison: Mocap vs LRF

Condition	Mean Mocap (cm)	Mean LRF (cm)	RMSE
50cm step length	49.1 $\pm$0.9	47.0 $\pm$3.9	4.1
25cm step length	24.8 $\pm$1.9	27.7 $\pm$3.4	4.3
free step length	46.9 $\pm$4.5	46.0 $\pm$4.2	3.2

The load data from the force plates and from our proposed device were trimmed for capturing only the walking part and the RMSE was obtained. The results were compiled in table II.

TABLE II
Vertical GRF comparison: Force Plate vs Load-cells

Condition	Mean Force Plate	Mean Load-cells	RMSE (Kg)
25% BWS	56.0 $\pm$4.6	57.3 $\pm$4.2	2.31
50% BWS	57.0 $\pm$5.2	58.6 $\pm$2.8	3.36
75% BWS	27.6 $\pm$2.2	28.8 $\pm$2.5	2.88

The symmetry index was calculated for each gait cycle using the vertical loads of the left and the right feet. Figure 3 shows the box-plot of the indexes of four participants in the condition of walking freely and walking with the orthosis (Right orthosis: P1,P2 Left orthosis: P3,P4).

Finally, the average and the STD of the GRF during normal walking were calculated. Figure 4 shows the average on GRF supported by each limb grouped by the amount of BWS.

V. Discussion

The error comparing the proposed device with the motion capture presented similar values independent of the step length (table I). Considering the variability of natural gait and the purpose of this study, as to provide a metric for the evaluation gait rehabilitation, this may be considered acceptable as an estimation. Regarding the sensor configuration, leg occlusions on the LRF information were noticed in some specific conditions. Despite not representing a problem for the gait parameters described in this paper, it may compromise further development which requires monitoring during all gait cycles. To overcome this condition an additional LRF can be used and its data can be fused to avoid occlusions.

In table II, the vertical GRF measured by the platform has values lower than the acquired by the force plate. Since the

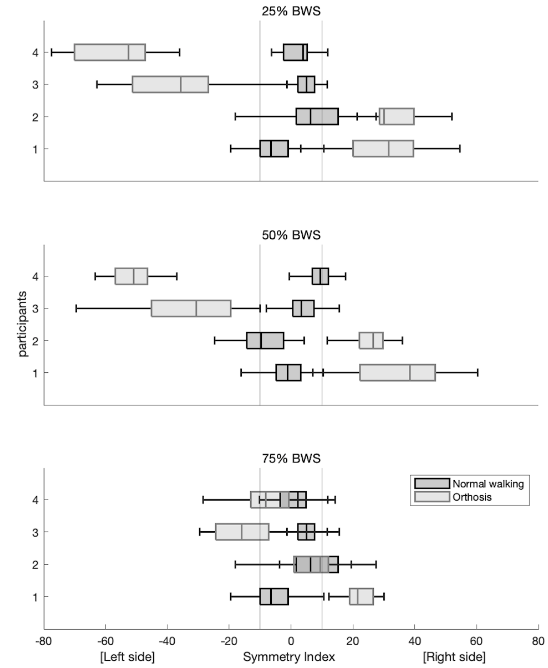

Fig. 3. Gait Symmetry Index based on GRF for four participants. (Part. 1 and 2 right orthosis. Part. 3 and 4 left orthosis.)

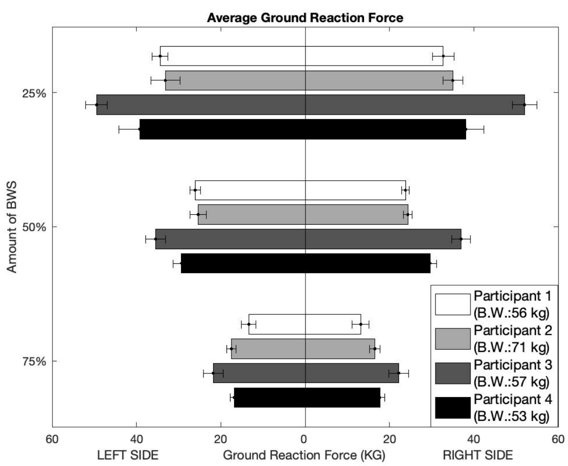

Fig. 4. Average on Ground Reaction Force (GRF) on the left and right leg of participants with different body weight unloading.

peaks in the force of both sensors were acquired automatically, we hypothesize that since the capture frequency of the force plate is much higher than our circuit, it could detect the deacceleration of the limbs when hitting the ground. Moreover, more careful calibration of the load-cells may contribute to stabler results.

The symmtery index observed for the 25% and 50% BWS conditions (Figure 3) showed that participants were leaning towards the side of their body with the orthosis. However, in the 75% BWS condition, participants were more symmetry. We hypothesize that this might be due to participants being unable to have proper ground contact with their feet because of the high amount of body weight unloading. This lack of ground contact, and small value of the calculated GRF might give the impression of symmetry. However, this phenomena is not investigated in this study and would require further studies to clarify the factors contributing to this supposed gait symmetry.

The instantaneous total GRF is presented in a display in front of the therapist throughout the session. It can be useful to fine adjustment of the body weight unloading according to different participants. Furthermore, the average in the lateral GRF was calculated for each session, which may provide a metric to estimate the muscular strength performed by participants during walking.

VI. CONCLUSIONS

A new method for measuring gait information of participants during walking rehabilitation is proposed in this paper. By comparing the error with traditional methods of measurement, we believe that our device can be effective for fast evaluation of individuals without direct attachment of sensor. Also, we extended the use of the sensor with metrics for gait asymmetry and provided further information, such as amount of body weight support and ground reaction forces. It can be beneficial to therapist for a more personalised treatment of patients. For further works, we plan to use the acquired gait information to dynamically control the body weight support of participants during overground walking.

REFERENCES

[1] J.-M. Belda-Lois, S. Mena-del Horno, I. Bermejo-Bosch, J. C. Moreno, J. L. Pons, D. Farina, M. Iosa, M. Molinari, F. Tamburella, A. Ramos, A. Caria, T. Solis-Escalante, C. Brunner, and M. Rea, "Rehabilitation of gait after stroke: a review towards a top-down approach," *Journal of NeuroEngineering and Rehabilitation*, vol. 8, no. 1, p. 66, 2011.

[2] J. Nonnekes and A. Nieuwboer, "Towards Personalized Rehabilitation for Gait Impairments in Parkinson's Disease," *Journal of Parkinson's Disease*, vol. 8, pp. S101–S106, 2018.

[3] M. Leite, B. Soares, V. Lopes, S. Santos, and M. T. Silva, "Design for personalized medicine in orthotics and prosthetics," *Procedia CIRP*, vol. 84, pp. 457–461, 2019.

[4] A. Wagner, "A Rehabilomics framework for personalized and translational rehabilitation research and care for individuals with disabilities: Perspectives and considerations for spinal cord injury," *The journal of spinal cord medicine*, vol. 37, 2014.

[5] J. Marin Bone, J. Marin, T. Blanco, J. la Torre, I. Salcedo, and E. Martitegui, "Is My Patient Improving? Individualized Gait Analysis in Rehabilitation," *Applied Sciences*, vol. 10, p. 8558, 2020.

[6] J. Marin, T. Blanco, E. Martitegui, and J. C. Aragu, "Integrating a gait analysis test in hospital rehabilitation : A service design approach," pp. 1–23, 2019.

[7] S. A. Alves, R. M. Ehrig, P. C. Raffalt, A. Bender, G. N. Duda, and A. N. Agres, "Quantifying Asymmetry in Gait: The Weighted Universal Symmetry Index to Evaluate 3D Ground Reaction Forces," *Frontiers in Bioengineering and Biotechnology*, vol. 8, p. 1189, 2020.

[8] T. Ramakrishnan, C.-A. Lahiff, and K. B. Reed, "Comparing Gait with Multiple Physical Asymmetries Using Consolidated Metrics," *Frontiers in Neurorobotics*, vol. 12, p. 2, 2018.

[9] B. Leme, M. Hirokawa, H. Kadone, and K. Suzuki, "A Socially Assistive Mobile Platform for Weight-Support in Gait Training," *International Journal of Social Robotics*, 2019.

[10] M. Mcgrath, D. Howard, and R. Baker, "The strengths and weaknesses of inverted pendulum models of human walking," *Gait & Posture*, vol. 41, 2014.

[11] W. Herzog, B. Nigg, L. Read, and E. Olsson, "Asymmetries in ground reaction force in normal human gait," *Medicine and science in sports and exercise*, vol. 21, pp. 110–4, 03 1989.