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Portable electronic device to assess the human balance using a minimum number of sensors

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Portable electronic device to assess the human balance using a minimum number of sensors

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Abstract

Objective: This paper presents the design and development of a new electronic portable device to assess the human balance of the human body during standing, using a minimal number of sensors and peripheral components. This device is aimed to evaluate human balance in environments outside of specialized laboratories, such as small clinics and therapy offices.

Approach: The design is based on previous designs using three or more resistive force sensors attached to the feet, however in the present work, the sensors were attached on an adjustable platform, to fit several sizes of feet. Furthermore, all the signal acquisition, process, storage and display are executed by an embedded electronic system, thus avoiding the use of computers and external peripherals. A new method to compute the CoP using only two sensors per foot was developed and tested in a group of 50 university students, (17 women and 33 men), 26.04±4.94 years.

Main results: It was developed a portable electronic system to measure the trajectory of the CoP and to calculate the indexes values derived from it. The system is capable to discriminate between measuring situations (open and closed eyes), using only two sensors per foot ($p<0.0001$). A comparison between the values obtained for young subjects using the proposed device, and the values reported in the literature showed a similar tendency.

Significance: The results indicate that the proposed system is a good, low-cost, and easy-to-use alternative tool for researchers and clinicians interested in the evaluation of human balance, especially if the measurements must be done outside laboratories.

Keywords: Human balance, portable system, minimum points of sensing, Centre of Pressure trajectory.

1. Introduction

The human balance contributes greatly to the quality of life, since it allows to execute very important tasks for autonomy, such as standing, walking or running. Physiologically, the balance is controlled by the central nervous system (CNS), which in turn, activating the musculoskeletal system, compensates the position of the center of mass (CoM) (Winter 2009). The dependence of the balance on important physiological systems (visual, vestibular, somatosensorial, neurological and musculoskeletal) make of the balance assessment an important tool to evaluate indirectly, the health status of these systems (Winter 1995). Examples of the usefulness of balance measurement to evaluate disorders can be found elsewhere, for example, the effects of using high heels (Cho and Choi 2005), the balance and gait capability in survivors from strokes (Rodgers *et al* 2004), the effects of neurological disorders in elderly (Escudero *et al* 2013), effects of diabetes in stability (Lin *et al* 2012), effects of age in balance (Berg *et al* 1992), etc. The most common method to evaluate the human balance is measuring the center of pressure (CoP) of the body (Winter 1995), during static posturography (Janusz *et al* 2016). The CoP represents the weighted average of the body pressure exerted by the feet surface to the ground. The sway amplitude of people standing upright, modify the plantar pressure exerted on the contact surface, thus, measuring those pressure changes in the sagittal plane (Antero-Posterior -AP-) and the frontal plane (Medio-Lateral -ML-) it is possible to evaluate quantitatively the balance. The graph of displacement signals in AP or ML versus time, it is known as stabilogram (Winter 1995, Perry 2010), while the graph obtained from the ML versus AP signals it is called statokinesiogram (Perry 2010), which reflects the trajectory of the CoP.

There are several indexes derived from the stabilograms and statokinesiogram, which allow to assess the balance quantitatively (Morasso *et al* 1999). Among the most reported indexes are: the mean, RMS value, velocity of the main component distance, main frequencies of the trajectory of CoP, and the area of the statokinesiogram circumscribed to a circle or an ellipse (Prieto *et al* 1996, Duarte and Freitas 2010). Due to the high variability inter and intra subject, as well as the variability on the devices used to measure the CoP and the lack of standards on the test conditions (Tamburella *et al* 2014), it has been not possible to reach an agreement on the standardization for the values of the indexes (Cornilleau-Pérès *et al* 2005), (van der Kooij *et al* 2011), (Janusz *et al* 2016) (Koltermann *et al* 2019). For this reason, is not possible to compare the values obtained by a given subject to a universal standard. The most common assessment method is to measure the CoP for each subject under different conditions, one of them more challenging for the equilibrium than the other and to compare these results. To date, the most used method to assess the balance in static posture is the Romberg´s test (Cornilleau-Pérès *et al* 2005, García-Pastor and Álvarez-Solís). This test consists on the measurement of the CoP during two conditions: open eyes and closed eyes. The posture for the test consists in the subject, standing still, feet together, crossed arms over the chest and looking at the front. From this evaluation is calculated the Romberg´s quotient (the division of the values obtained during the closed eyes condition by the values obtained during the open eyes condition). This quotient is usually greater than one and the magnitude will depend on the impairment level of the balance system, however, as mentioned before, there not exist standardized values.

The Romberg´s test has been successfully used to identify several causes of balance disorders, such as inherited disorders, toxic and metabolic disorders, immunological diseases, and neurological disorders (García-Pastor and Álvarez-Solís 2013).

The most common devices that generate the statokinesiograms and compute the indexes used in the Romberg´s tests are the force platforms (Faulkner and Robinson 1996), stabilometers (Escudero et al., 2013) and customized instrumented insoles (Winter 2009, Abou Ghaida *et al* 2014). Force platforms and stabilometers are considered the gold standard in CoP measurement (Rocchi *et al* 2004) because these devices can register all forces and moments exerted on their contact surface. The main disadvantages of the force platforms are the low portability and high cost, which limits their use to specialized clinics and laboratories (weight around 90 kg and cost around $20,000). Thus, for clinical research or routine screening it is necessary for the patients to displace to the laboratory. However, it is obvious that for patients suffering balance disorders is not easy to move from his/her home/ asylum or hospital to the laboratory facilities. Thus, hampering the use of the balance assessment as a tool for routine health checks or follow-up treatments.

Trying to cope with these limitations, several alternatives have been proposed, such as the use of commercial portable stabilometers (Biodex®, New York, USA), and instrumented insoles (Philip S Dyer 2011, Abou Ghaida *et al* 2014, Nagamune and Yamada 2018), trying to make devices more portable, low cost and easy to deploy outside laboratory facilities. However, all of these alternatives still have some drawbacks that can be improved.

For example, the commercial portable stabilometer, made by Biodex.inc, cost around 8,000 USD and weights around 20 kg (including the case to carry it). Its cost and limited portability are still a considerable obstacle to use it in small clinics or therapy offices.

Aiming to reduce these disadvantages, it has been proposed the use of the Wii Balance Board (WBB), a force platform designed and manufactured by Nintendo® (Kyoto, Japan) intended as an interface for videogames. There are plenty of reports claiming the usefulness of the WBB to measure the CoP as good as the force platforms do (Clark *et al* 2010). The characteristics of the WBB, such as its low cost and lightweight device (around 90 USD and 3.5 kg) as well as its proved reliability and repeatability (Clark *et al* 2018), make of this device an interesting alternative to measure the CoP at low cost and in environments outside laboratory.

However, there are at least two important drawbacks for this device that need to be solved, before this platform becomes a serious alternative. First and most important, the data acquisition from the WWB is not reliable. The jitter of the device is as large as 60 ms (Pagnacco *et al* 2011, StackExhange 2014, Goble *et al* 2014). Thus, the sampling rate is not constant at all (vary from ~30 to 70 Hz), something very important for the analysis of the signal, especially for frequential indexes. The average sampling rate of the device is around 63.6 Hz and is not possible to be controlled by the host PC user. In addition, there are not measures to avoid repeated or lost samples. This is good for the device used as a videogame interface, but not for a clinical device (Pagnacco *et al* 2011). A signal with data losses and highly variable sampling rate, lead to flawed results (Leach *et al* 2014, Audiffren and Contal 2016). The second disadvantage is the necessity of a dedicated computer (to reduce the data loss) to interface the device, thus increasing the cost and complexity of the system and reducing its portability.

For the case of instrumented insoles, measurement of pressure attaching force sensors on the foot have proved to be reliable and consistent with measurements done by force platforms (Philip S Dyer 2011, Abou Ghaida *et al* 2014, Nagamune and Yamada 2018). These systems employ several sensors per foot, which must be fixed manually for each subject, depending on the foot size. Furthermore, the type of sensor employed (FSR, interlink electronics®) tend to wear and tear when manipulated frequently, thus forcing to change sensors often.

The commercially available insoles, such as Open Go (Moticon Science Inc., Germany) cost around 1, 900 USD a price relatively high considering that they are designed and must be customized for a unique user in order to function properly. Other insoles, intended for research purposes, such as F-SCAN system with instrumented insole (Tekscan, inc., USA) or NOVEL (Pedar, Germany) use many sensors (from 85 to 955 per insole) and require dataloggers attached to the subject, only to acquire the data. Furthermore, their prices are prohibitive for small clinics or therapist (around 10,000 USD) plus 35 USD per insole.

In this paper is presented a system to evaluate the CoP, using only two force sensors per foot attached to an adjustable platform. Aiming to offer an alternative to evaluate the balance at a low cost, with a portable system that allows to assess the balance in small clinics or places where the patient´s activity takes place, such as asylums, elderly clubs, etc.

The system presented does not require a computer to acquire, processing and showing the results, thus reducing cost, increasing portability and the user-friendliness.

The system presented is based on that described in (Abou Ghaida *et al* 2014) which uses three sensors per foot. In this paper, first were used three sensors per foot, in order to reproduce the results reported by Abou Ghaida. Once the system was tested using three sensors, an analysis using only the signals of two sensors per foot was developed. This analysis allowed to assess the capability of the system to measure the CoP trajectory using a minimum number of sensors per foot.

The acquisition, processing and computation of the CoP and stability indexes are executed in embedded electronic system, avoiding the use of external computers, as required by most of the commercial systems and those based on the WBB. A preliminary test to evaluate balance in young healthy subjects is also presented.

2. Materials and methods

2.1 System development

To determine the number and position of the sensors to use, it is important to know the most adequate points to fix the sensors. (Cavanagh *et al* 1987) determined that for young healthy adults between 20 to 40 years old, the distribution of plantar pressure when standing in a flat surface, is as follows: 60.5% of body weight is supported by the heels (211.75 N), 28.1% in the metatarsal region (98.35 N), 7.8% by the middle foot (27.3 N), and 3.6% by the fingers (12.6 N), see Figure 1. The forces exerted by the metatarsals and heels to the ground represents around the 88% of the total force.

Figure 1. Distribution of plantar pressure.

Several approaches have been developed to measure these forces. The works reported by (Philip S Dyer 2011, Nagamune and Yamada 2018) employed 10 sensors per foot, with results comparable to a force platform. (Abou Ghaida *et al* 2014) demonstrated that using only three sensor per foot, it is possible to calculate the CoP similarly to a commercial system (F-SCAN® system from Tekscan.inc).

The work done by (Abou Ghaida *et al* 2014) employing 3 sensors per foot was replicated and improved in this research. By developing a movable platform that allows adjusting the position of the sensors depending on the size of the feet of the subjects, instead of attaching the sensors to the feet directly, (see in Figure 2 the position of each sensor label S1 to S6). The sensors employed are the same as those used by (Abou Ghaida *et al* 2014), (FSR 402 and FSR 406, from Interlink. Inc). These sensors are thin, low-cost, small, light, and their associated instrumentation is easy to implement. The thinness of this type of sensors reduce discomfort when stepped or standing on them. Two FSR 402 sensors were used to measure the force on the metatarsal region and one FSR 406 for the heel region. These sensors have already used successfully in similar applications (Rafajlović *et al* 2009, Philip S Dyer 2011, Abou Ghaida *et al* 2014), reporting reliability and effectiveness, even in walking tests.

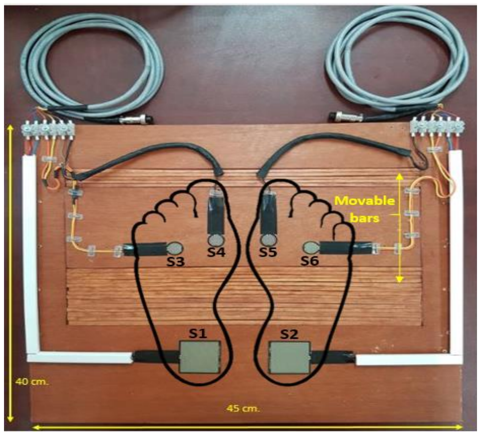

Figure 2. Platform with sensors and proposal press.

In order to determine the sampling frequency and amplification for the signal acquisition, it was considered that the frequency content of the signals for healthy adults, is in the range from 0.1 Hz to 2 Hz (Fujimoto *et al* 2014), and for elderly patients with some vestibular dysfunction is in the range from 0.1 Hz to 5 Hz. The amplitude of the CoP displacements in the AP and ML is lower than 5 cm (Rocchi *et al* 2004). Considering this, to obtain reliable data, it is recommended to record at least 10 or more periods of the lowest frequency from the signal of interest (Grimaldi and Manto 2012), and sampling at more than 5 or 10 times the highest frequency, then the system should be capable to sample at least at 25 Hz and to record at least 10 seconds of the signal from the balance test. Usually, a common balance test has a duration of 30 seconds as minimum (Prieto *et al* 1996, Latash *et al* 2003) with sampling rates from 25 to 100 Hz.

The separation between feet and thus the attaching place of the sensors is based on the reports from (Lawrence and Schmidt 1997), and (Cruz *et al* 2010) which state that in order to evaluate the balance and postural stability, the subjects must assume positions that demand greater efforts to keep the stability. From the four positions that (Cruz *et al* 2010) exposes, it was chosen the position with feet together, because is the one that most compromises the postural balance, improving the detection of balance disorders.

To make the platform adjustable to any length of foot, a set of movable bars were implemented on it (0.5 cm wide each one, see Figure 2). By changing the position of the bars, it is possible to move the base containing the metatarsal sensors and, at the same time, maintain the same level of contact surface for the entire platform. The sensors to acquire the force exerted by the heels remain fixed on the platform.

A voltage divider circuit, followed by a low pass passive filter (cutoff frequency of 7 Hz) were used to conditioning the signal before to digitalize it (see Figure 3). The component values were chosen in order to obtain a voltage signal ranging from 0 to 5 V. The RC filter is used as an anti-alias filter before digitalizing. The conditioned signals are sampled at 100 Hz using the 12-bit ADC of the dsPIC30F6014A (Microchip^{TM}).

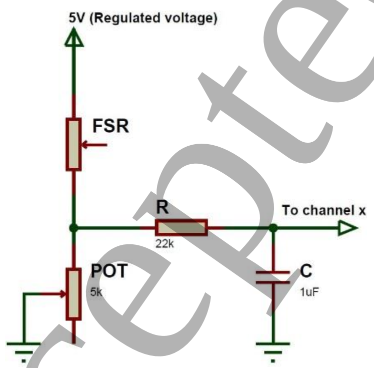

Figure 3. Signal conditioning for each FSR sensor.

The dsPIC30F6014A also manage the recording the acquired data on a micro-SD card and drives the interface of the system based on a TFT touch screen display. The Figure 4 shows the electronic board for the interface and data save module of the system. The microcontroller, connection terminals for the filtered signals and the regulated voltage are on back side of this electronic board. Four LEDs were included in this design to indicate about the running tasks, such as sampling, writing/reading in SD or absence of it in the system.

Figure 4. Electronic card of digitization stage.

The electronic boards and the battery for power supply are embedded in an ABS plastic case made using a generic 3D printer. Figure 5 shows the system placed in the case displaying the initial menu, ready to acquire CoP signal.

Figure 5. Embedded signal processor system.

2.2 Characterization and calibration of the system response

In order to guarantee that the signals acquired by the sensors correspond to the force applied, a calibration process for each sensor was developed. A dynamometer AFG100 (Mecmesin®) was used to measure the applied force. This device was adapted to a bench to apply vertical force using a crank and a screw (see Figure 6 (a)).

A rubber surface was attached to the tip of the dynamometer in order to avoid the direct contact between the sensor and metal surface of the device, trying to mimic the contact of the foot with the sensor (see Figure 6 (b)).

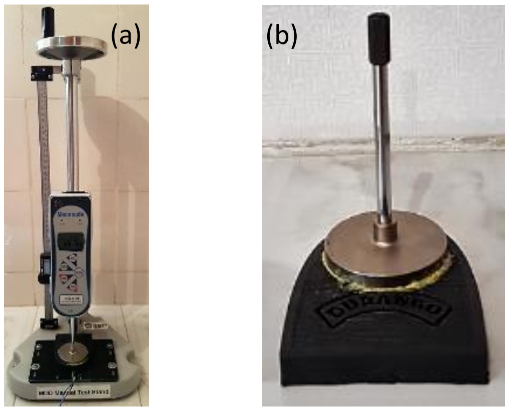

Figure 6. Elements for characterization of the sensor, (a) Dynamometer with bench and (b) Pressurizing rubber.

For each sensor, a force from 1 to 115 N (in steps of 5 N) was applied during periods of 20 seconds, with the purpose of stabilize the sensor response. This calibration process was determined heuristically after to make several trials of characterization. The signals obtained from this process were conditioned, digitized and stored by the developed system. Subsequently, the software MATLAB® 2015B (MathWorks, Inc) was used to analyze the data. A curve adjustment was made by polynomial interpolation for each sensor. After trying several curve fittings models, a third-order polynomial resulted the best option. The approximation polynomials obtained for each sensor are the equations 1 to 6 respectively.

$$S1 = 5.097x10^{-6}(X_1{}^3) - 9.47908x10^{-4}(X_1{}^2) + 0.183309138(X_1) - 2.595683384 \quad (1)$$

$$S2 = 2.645x10^{-6}(X_2{}^3) - 3.67880x10^{-4}(X_2{}^2) + 0.144392112(X_2) - 1.054611716 \quad (2)$$

$$S3 = 3.294x10^{-6}(X_3{}^3) - 6.92254x10^{-4}(X_3{}^2) + 0.179867558(X_3) - 1.550111898 \quad (3)$$

$$S4 = 3.552x10^{-6}(X_4{}^3) - 3.64081x10^{-4}(X_4{}^2) + 0.120829087(X_4) - 1.434386736 \quad (4)$$

$$S5 = 2.550x10^{-6}(X_5{}^3) - 4.78195x10^{-4}(X_5{}^2) + 0.185680929(X_5) - 2.222566799 \quad (5)$$

$$S6 = 2.754x10^{-6}(X_6{}^3) - 4.58775x10^{-4}(X_6{}^2) + 0.154582956(X_6) - 1.623776455 \quad (6)$$

Where:
S1 to S6: are the forces applied to each sensor respectively.
$\boldsymbol{X_n}$: is the digitized value of each sensor (n from 1 to 6).

The comparative graph of the applied force and adjusted data is shown in Figure 7. The continuous line represents the values calculated from the equations 1 to 6, while the "*" represents the measured values for each force applied. To evaluate the quality of the adjusted polynomials, the Pearson's correlation coefficient was calculated for each one. All the graphs presented an excellent correlation value (the minimum was 0.933, with p<0.001). Once the response of the sensors was characterized, the polynomials were programmed into the microcontroller.

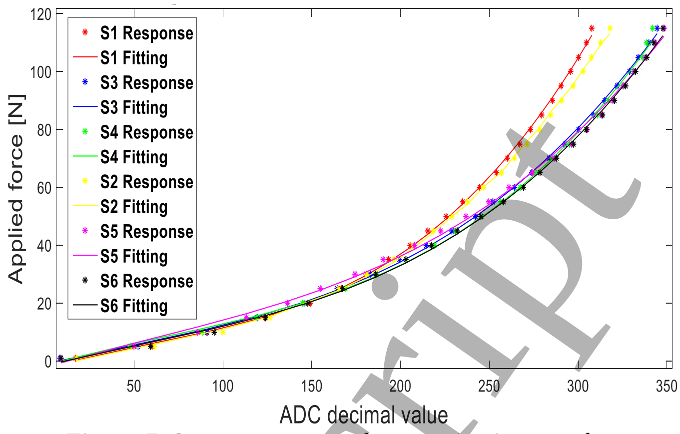

Figure 7. Sensor response characterization graph.

2.3 Calculating the center of pressure using three sensors

To compute the CoP using three sensors per foot, an adaptation of the work presented by (Huang *et al* 2013) was done. It is based on four measured points on a force plate. A rectangle is assumed considering the distance between the sensors placed on the region of the heels and the distance towards the location of the third metatarsal, the area of the rectangle "ab", as shown in the Figure 8.

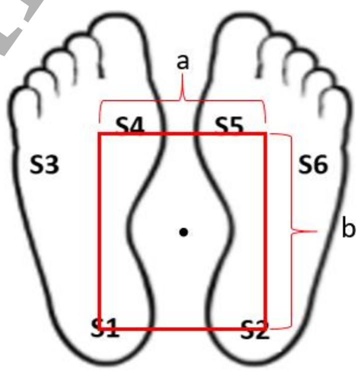

Figure 8. Graphical adaptation to a force platform.

The calculation of the CoP in A/P and M/L directions is done using the equations 7 and 8 respectively, which result from an adaptation of the equations reported by (Huang *et al* 2013).

$$COP_X = \frac{a}{2}\left(\frac{(S5+S6+S2)-(S3+S4+S1)}{S1+S2+S3+S4+S5+S6}\right) \quad [\text{cm}] \qquad (7)$$

$$COP_Y = \frac{b}{2}\left(\frac{(S3+S4+S5+S6)-(S1+S2)}{S1+S2+S3+S4+S5+S6}\right) \quad [\text{cm}] \qquad (8)$$

Where:
$\boldsymbol{COP_X}$: CoP displacement in M/L or frontal plane.
$\boldsymbol{COP_Y}$: CoP displacement in A/P or sagittal plane.

S1 to S6: Sensed forces and adjusted using equations 1-6.
a: distance between the central point of sensors S1 and S2.
b: distance given by the central point of S1 (or S2) and the geometrical middle point between S3&S4 (or S5&S6 if using S2)

For the built platform, distance "a" is 23 cm according to (Abou Ghaida *et al* 2014). Distance "b" will change according to the size of the feet. The fixed value of "a" distance limits the use of the platform to adult patients who can assume the test position. However, this position is one of the most used for the Romberg´s test, so, the applications of the device are still many. However, the distance "a" can be easily modified in the design, depending on the type of subjects to be measured.

The distance "b" should be registered manually in the system once adjusted.

A. Signal Processing

In order to eliminate the DC offset given by the force exerted by the body without movement, the mean value of each signal was subtracted by applying the equations (9) and (10).

$$CoP_{-offsetX}\,[i] = COP_X[i] - \overline{COP_X} \quad [\text{cm}] \qquad (9)$$

$$CoP_{-offsetY}\,[i] = COP_Y[i] - \overline{COP_Y} \quad [\text{cm}] \qquad (10)$$

Where:
i: number of the sampled data (from 0 to the total number of samples).
$CoP_{-offsetX}$: M/L CoP displacement without offset.
$CoP_{-offsetY}$: A/P CoP displacement without offset.
$\overline{CoP_X}$ and $\overline{CoP_Y}$: mean of the CoP calculated signals.
$CoP_X[i]$ and $CoP_Y[i]$: each of the samples of CoP.

The most common indexes reported for balance evaluation are calculated from the CoPx and CoPy signals. These indexes are shown in Table 1. The indexes description and computation formulas are reported elsewhere (Prieto *et al* 1996, Qiu and Xiong 2015).

Table 1. Indexes calculated from the CoP displacement signal.

Main index	*Derived index*	*Total number of indexes*
Resultant distance (RD)		1
Mean distance (MDIST)	MDIST-AP MDIST-ML	3
Root mean square distance (RDIST)	RDIST-AP RDIST-ML	3
Total of excursions (TOTEX)	TOTEX-AP TOTEX-ML	3
Mean velocity (MVELO)	MVELO-AP MVELO-ML	3
Excursion range of the CoP (RANGE)	RANGE-AP RANGE-ML	3
95% confidence circle area (AREACC)		1
95% confidence ellipse area (AREACE)		1
sway area AREASW		1
Mean frequency (MFREQ)	MFREQ-AP MFREQ-ML	3
Fractal dimension based on circle (FDCC)		1
fractal dimension based on ellipse (FDCE)		1
	Total number of indexes	**24**

All calculations and procedures are executed into the dsPIC and stored in the SD card in a new different file. For each measurement, the system stores three files: one file for the raw sensor signals, the second file for the CoP signals, and the third file for the calculated indexes values.

B. Measurement comparison of CoP signals

In order to compare if the acquired data correspond to the CoP signals measured with another device, it was implemented a tests using a WBB platform. It was desirable to use a commercial platform to measure the CoP, however our laboratory does not have any, so a WBB was used.

As (Clark *et al* 2010) mentioned, the WBB is capable to measure the CoP as well as a laboratory platform. However, as mentioned in the introduction this device has the disadvantage of an unstable sampling rate. In order to avoid this disadvantage, the WBB electronic board was modified to guarantee a fixed sampling rate and thus to avoid this limitation of these type of devices.

For the comparison test, the proposed system was placed on the WBB surface (as shown in the Figure 9) and simultaneous measurements were made. One subject stand on the devices for 20 seconds, swaying on purpose, in order to compare the resulting CoP signals.

Figure 9. Simultaneous CoP measurement between proposed system and WBB.

The Figure 10 and Figure 11 show the resulting signals. Although the signals are not completely similar, a correlation analysis resulted in a 93.3 % of similarity for the A/P direction and 86.4% in the M/L direction, which is considered as a good similarity value.

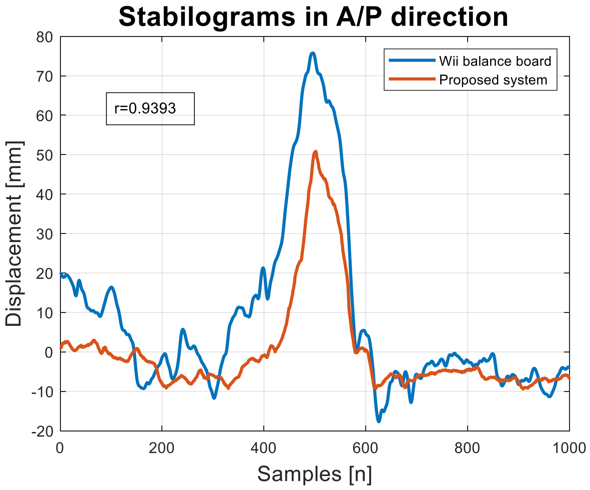

Figure 10. A/P stabilograms by system and WBB.

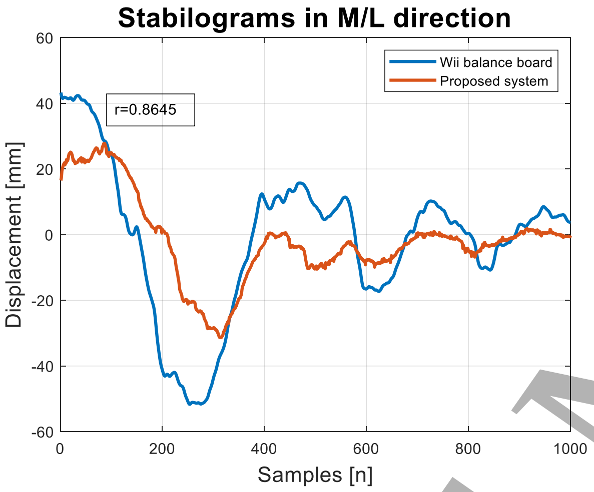

Figure 11. M/L stabilograms by system and WBB.

C. Development results

The Table 2 presents the main features of the developed system. The construction cost at prototype level is around $500 USD. Although the prototype could not be compared directly with commercial systems in cost, this system represents a portable alternative tool to quantify the center of pressure, with three sensors.

Table 2 System characteristics summary.

Weight	Electronic system: 660 g. Platform: 1.170 kg.
Size	Electronic system: 15x20x5 cm. Platform: 45x40 cm.
Measurement range	-5 to 5 cm in A/P & M/L
Control interface	Touch screen.
Evaluation time per test	30 seconds.
Sampling frequency	100 Hz.
Power supply	5 V DC & 350 mA.
Battery attributes	7.4 V DC & 1000 mAh. System autonomy: 2 hours. Recharge time: @ 90 minutes.
Storage capacity	SD card form 4 GB to 32 GB

Considering the size, weight and autonomy of the final prototype, it can be inferred that the proposed system rivals with the other similar systems reported in the state of the art(Dyer and Bamberg 2011, Abou Ghaida *et al* 2014) and even with commercial systems such as the Biosway® (Biodex).

2.4 Tests protocol

In order to evaluate the performance of the system measuring balance, a total of 61 university undergraduate and postgraduate students between 20 and 39 years old participated in a test. Considering the main criteria of inclusion, the absence of recent injuries or fractures (in the last year), 50 students were selected, 17 women and 33 men, 26.04 ± 4.94 years old, 168.94 ± 6.13 cm height, 68.37±8.15 kg weight and 25.79 ± 0.97 cm in foot size. To minimize possible variables that could affect the static balance, participants were previously asked not to drinking alcohol, coffee or tea at least during the morning of the day of the test, furthermore, sleep at least eight hours the night before the test. Prior to the evaluation, each person signed a letter of consent and afterwards was asked to answer a questionnaire related to the frequency of consumption of alcohol, smoking, fractures/injures, sleep cycle, chronic pathologies and drug management.

The test protocol is based on those presented by Vilma Gonzalez, Ma (Vilma Ivania Keglevic Román 2004, Ma *et al* 2014) and Norris (Norris *et al* 2005). The most common activity in those works is the realization of the Romberg's test (da Silva *et al* 2012, García-Pastor and Álvarez-Solís n.d.), in which the balance of the subject is evaluated under two condition: open eyes and close eyes, both conditions with feet together, crossing the arms and with the palms of the hands touching the shoulders. It is also suggested that the evaluated subject wears light clothes (short and t-shirt). The test protocol is as follows:

1. The person is asked to stand on the instrumented platform guaranteeing that the heel and metatarsal area are in contact with the sensors.
2. The person must assume an upright position as still as possible as indicated by the Romberg test (Ma *et al* 2014).
3. Maintaining the aforementioned position, the person must look at a bullseye placed in front of him/her, located at eye level and at a distance of two meters; focusing his/her attention on it, during the data acquisition. The Figure 12 shows a participant during the test, assuming the required position.

Figure 12. Posture for the balance evaluation.

4. The following test (closed eyes) is performed subsequently, without the subject stepping down from the platform, with the purpose of keeping the sensors located exactly as for the previous test (open eyes).

Both conditions (open eyes/closed eyes) are maintained for 30 seconds each one.

3. Data analysis and results

Considering the inclusion criteria mentioned above, only the measurements of fifty students were considered for the analysis of signals. For each subject, the system generated six files (raw sensor signals, signals without static component and indexes) for each vision condition. Figure 13 show a picture of the system display drawing the statokinesiogram generated by a 26-year old male, 167 cm of height, 69 kg of weight and 26 cm foot size with open eyes. In the Figure 14 are shown the results for the same subject with closed eyes. It can be observed a wider CoP excursion in both directions compared to Figure 13 this due to the vision privation. With closed eyes, the body sway is more noticeable in this phase, and therefore, the value of all CoP indexes increases.

In order to determine if the signals acquired by the system are useful to discriminate between open and closed eyes tests, a posterior analysis was carried out using a script developed in MATLAB® 2015B (MathWorks, Inc). All the signals were previously filtered, using a digital low-pass filter at 7 Hz cutoff frequency (fourth order Butterworth, Infinite Impulse Response). Then CoP displacement and indexes were calculated.

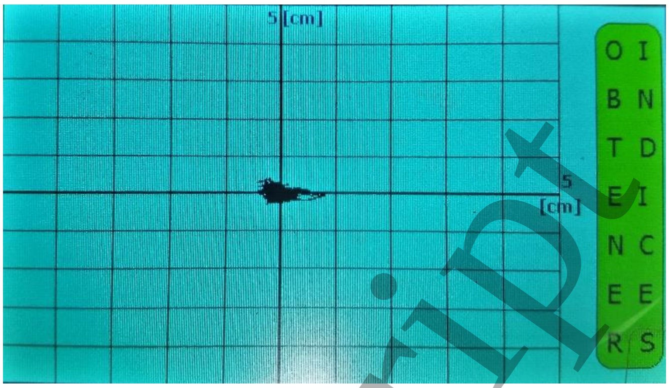

Figure 13. Statokinesiogram generated with open eyes.

Figure 14. Statokinesiogram generated with closed eyes.

3.1 CoP Index values from six sensors (three per foot)

Although the system can compute 24 indexes, for this analysis, only the 19 most commonly reported were included. The Table 3 shows the average and standard deviation of the index's values considering the six sensors of the platform.

Table 3. Comparison of indexes obtained with open/closed eyes.

Index	Units	Open Eyes	Closed Eyes	p value
MDIST	mm	1.71±0.33	2.66±0.52	*
MDIST-AP	mm	1.12±0.32	1.72±0.53	*
MDIST-ML	mm	1.04±0.29	1.66±0.59	*
RDIST	mm	1.97±0.36	3.10±0.61	*
RDIST-AP	mm	1.42±0.40	2.13±0.74	*
RDIST-ML	mm	1.30±0.34	2.12±0.65	*
RANGE-AP	mm	7.88±2.28	11.4±3.88	*
RANGE-ML	mm	6.43±1.76	9.79±3.06	*
MVELO	mm/s	5.50±0.97	6.47±1.33	*
MVELO-AP	mm/s	4.53±0.80	5.14±0.97	0.0004
MVELO-ML	mm/s	4.26±0.58	4.88±1.07	*
AREA-CC	mm2	70.96±26.77	181.70±78.36	*
AREA-CE	mm2	69.94±25.44	164.07±59.25	*
|	mm^2/s	2.99±1.06	5.50±1.66	*
MFREQ	Hz	0.39±0.09	0.51±0.09	*
MFREQ-AP	Hz	0.51±0.13	0.59±0.11	*
MFREQ-ML	Hz	0.60±0.19	0.76±0.24	*
FD-CC		1.08±0.03	1.12±0.026	*
FC-CE		1.08±0.03	1.13±0.026	*

* $p<0.0001$, with 95% confidence level.

It was proved by means of a paired t-test that there is a difference between the means obtained in test with open eyes and the means obtained with closed eyes for each calculated index.

Almost all the indexes obtained are considered statistically significant ($p<0.0001$), which implies that the system measurements are capable to detect changes in balance under two conditions.

D. CoP indexes values from four sensors (two per foot).

As stated in the introduction, the objective of this work is to reduce the number of sensors used to measure the CoP. In order to determine if the system can function using only two sensors per foot, the sensors S3 and S6 were eliminated from the calculus. These sensors were chosen because the force applied to them is slightly lower that for S4 and S5. By doing this, the signals of the CoP resulted slightly attenuated, so it was compensated in the computation of the CoP by multiplying the S4 and S5 values by a factor of two. Therefore, the equations 7 and 8 can be rewritten as show in equations 11 and 12.

$$COP_X = \frac{a}{2}\left(\frac{(2S5+S2)-(2S4+S1)}{S1+S2+2S4+2S5}\right) \quad [\text{cm}] \qquad (11)$$

$$COP_Y = \frac{b}{2}\left(\frac{2(S4+S5)-(S1+S2)}{S1+S2+2S4+2S5}\right) \quad [\text{cm}] \qquad (12)$$

The equations 9 and 10, used to eliminate the static component of signal remain unchanged.

To determinate the validity of this adjustment, a comparison between the CoP trajectory generated using six sensors and the CoP trajectory using four sensors was carried out. The similarity was assessed for each subject using the correlation function “*corrcoef*” included in MATLAB. The average for all the subjects, for open eyes condition in the M/L plane was 0.97± 0.02 and 0.95 ± 0.06 in the A/P plane; for closed eyes condition, the average for the M/L plane was 0.98 ± 0.02 and 0.96 ± 0.05 for the A/P plane. The p values were all below $p<0.0001$, which indicates an excellent correlation between both signals. The Figure 15 to Figure 18 show the comparison of the CoP trajectories with open and closed eyes in the A/P and M/L directions, for one subject, the correlation coefficient value (r) is also shown in the figures.

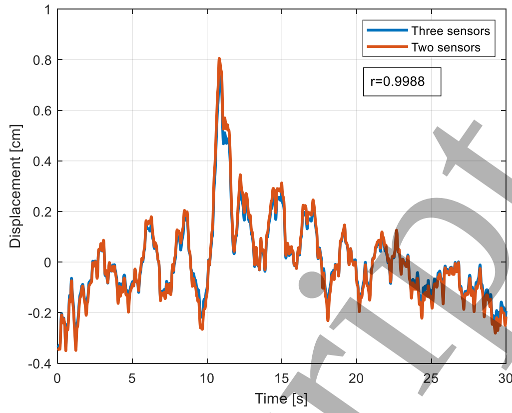

Figure 15. CoPx stabilogram comparison (open eyes).

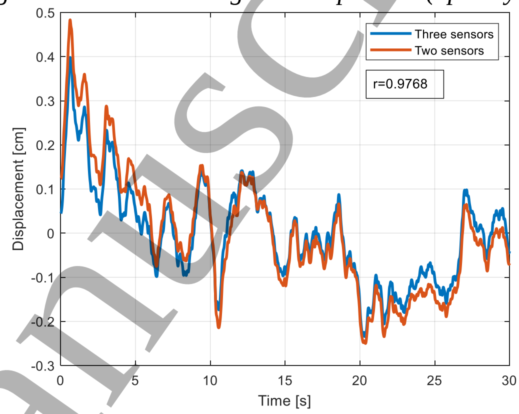

Figure 16. CoPy stabilogram comparison (open eyes).

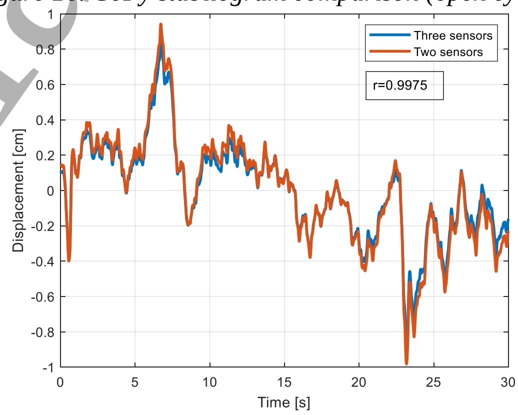

Figure 17. CoPx stabilogram comparison (closed eyes).

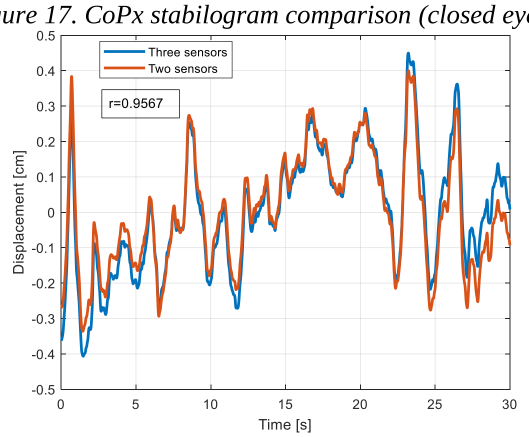

Figure 18. CoPy stabilogram comparison (closed eyes).

In order to observe the sway balance graphically and to make a qualitative comparison, the statokinesiograms were generated from previous stabilograms, the results are shown in Figure 19.

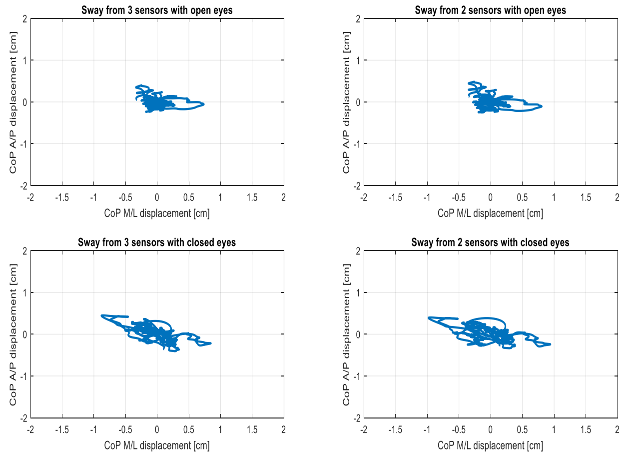

Figure 19. Generated statokinesiograms.

The correlation values indicate that the signals obtained using three or two sensors by foot are excellently related since the correlation factors are near to one and the *p* values are very small. To assess if these small differences could affect the capacity of the system to discriminate between open/closed eyes, the indexes were recalculated and tested again for each subject. The Table 4 shows the average values of these adjustments.

Table 4. Mean and standard deviations for open/closed eye tests using two sensors per foot.

Index	Units	Open Eyes	Closed Eyes	p value
MDIST	mm	1.80±0.42	2.78±0.62	*
MDIST-AP	mm	1.27±0.36	1.87±0.69	*
MDIST-ML	mm	0.99±0.35	1.67±0.58	*
R-DIST	mm	2.08±0.48	3.26±0.73	*
RDIST-AP	mm	1.61±0.45	2.38±0.85	*
RDIST-ML	mm	1.26±0.43	2.06±0.73	*
RANGE-AP	mm	8.83±2.49	12.77±4.35	*
RANGE-ML	mm	6.19±2.23	9.45±3.54	*
MVELO	mm/s	5.64±1.06	6.69±1.43	*
MVELO-AP	mm/s	4.79±0.90	5.49±1.14	0.0004
MVELO-ML	mm/s	3.12±0.61	4.74±1.06	*
AREA-CC	mm2	82.14±37.52	204.86±99.63	*
AREA-CE	mm2	77.90±35.48	178.32±70.99	*
AREA-SW	mm^2/s	3.09±1.28	5.73±2.01	*
MFREQ	Hz	0.39±0.09	0.51±0.12	*
MFREQ-AP	Hz	0.51±0.13	0.60±0.12	*
MFREQ-ML	Hz	0.57±0.18	0.71±0.22	*
FD-CC		1.08±0.03	1.12±0.03	*
FC-CE		1.05±0.03	1.08±0.03	*

* p<0.0001, with 95% confidence level.

According to results show in Table 4, it can be inferred that using only two sensors per foot the system is capable to detect significant differences between open/closed eyes tests as good as using three sensors. The p values of the paired t-test for all indexes were, by conventional criteria, statistically significant with a p<0.0001 value. Thus, any of these indexes can be used to discriminate balance for groups under different visual conditions.

Among the user tested during the develop of this project, there were two subjects, whose data were not included in the results showed in the Table 4 because they presented characteristics that did not fit in the groups of young healthy subjects. However, their results are presented here because are useful to emphasize the capabilities of the presented system to detect disorders in balance due either to age or muscle skeletal conditions.

The first case is a female subject 49 years old, apparently healthy, as mentioned by herself, however, she presented important changes on their signals and indexes values, compared to the young group. Her statokinesiogram is shown in the Figure 20 were is evident that she presented a larger sway on the ML plane compared to the younger subjects.

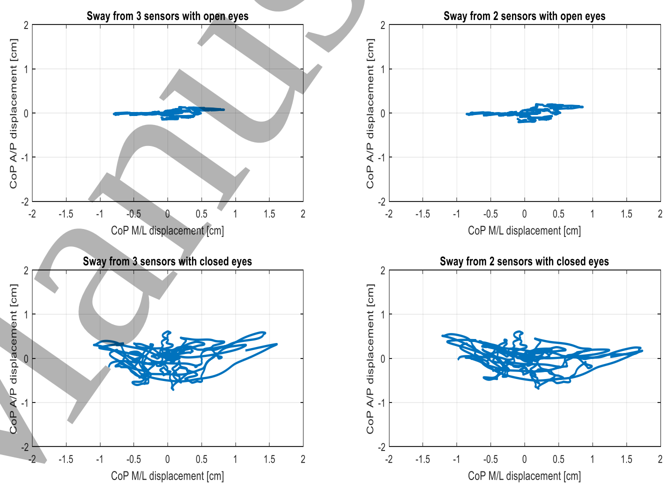

Figure 20. Statokinesiogram of a female subject, 49 years old.

Another interesting case corresponds to a female subject, 34 years old, who suffered a knee injury due to wear and tear of the joint on the left knee. She undergoes a surgery one year before, and the subject referred to feel completely recovered. However, her signals showed quantitative differences, compared to healthy subjects. Figure 21 shows her statokinesiograms using 2 and 3 sensors per foot.

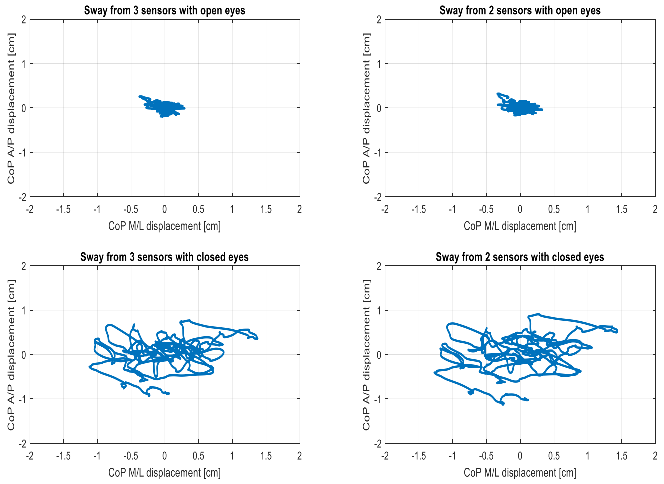

Figure 21. Statokinesiogram of a young female subject who suffered a knee surgery year before.

The Table 5 shows the comparison of the index's values among the healthy group and these two subjects with different characteristics. It can be observed that the indexes values show clear difference compared to the healthy and young group.

Table 5. Comparison of index values for healthy subjects vs subject with anomalies in the postural balance.

Index	*Romberg´s indexes for the healthy group* 3 sensors	2 sensors	*Subjects with differences to the healthy group* Knee subject 3 sensors	Knee subject 2 sensors	Aged subject 3 sensors	Aged subject 2 sensors
MDIST	1.59±0.39	1.62±0.51	4.39	4.8156	1.6518	1.6704
MDIST-AP	1.72±0.96	1.55±0.66	4.91	4.8518	1.3388	1.4850
MDIST-ML	1.68±0.69	1.88±0.11	3.70	4.6834	4.9763	3.0743
RDIST	1.61±0.39	1.63±0.51	4.43	4.7916	1.6383	1.6898
RDIST-AP	1.55±0.58	1.55±0.65	4.70	4.7850	1.4387	1.5761
RDIST-ML	1.74±0.84	1.83±0.16	3.96	4.8041	4.5696	2.9169
RANGE-AP	1.52±0.6	1.52±0.64	3.69	4.0454	1.6584	1.4609
RANGE-ML	1.63±0.71	1.66±0.82	3.76	4.1796	4.7288	3.2050
MVELO	1.20±0.25	1.21±0.27	2.21	2.2898	2.6548	2.7409
MVELO-AP	1.16±0.28	1.17±0.29	1.96	1.8514	2.0225	2.2852
MVELO-ML	1.14±0.21	1.15±0.21	2.22	2.5064	3.1663	2.7426
AREA-CC	2.84±1.81	2.98±1.43	19.9	22.7226	2.6509	2.9060
AREA-CE	2.57±1.19	2.7±1.57	18.65	22.9904	6.5772	4.5998
AREA-SW	2.00±0.79	2.08±0.99	2.45	2.1681	4.5975	3.7341
MFREQ	1.36±0.34	1.37±0.41	2.09	1.2831	1.6072	1.6408
MFREQ-AP	1.23±0.38	1.24±0.39	1.98	1.9777	1.5106	1.5388
MFREQ-ML	1.35±0.44	1.33±0.47	1.73	2.3767	1.4035	1.4926
FD-CC	1.04±0.03	1.04±0.04	1.10	1.1086	1.0672	1.0651
FC-CE	1.03±0.03	1.03±0.03	1.09	1.1050	1.0048	1.0683

Discussions

In this work is presented a portable electronic system to measure the CoP trajectory using only two sensors, which represents an advantage compared to similar works such as (Dyer and Bamberg 2011, Nagamune and Yamada 2018) which used 10 sensors per foot and (Abou Ghaida *et al* 2014) which used 3 sensors per foot.

This system also represents an advantage compared with other systems such as those based on WBB because it avoids the use of computers to acquire and analyze the data, thus reducing the cost and increasing the portability of the system. Furthermore, compared to WBB based system, which lacks from a steady sampling rate (Pagnacco *et al* 2011) and require a computer to acquired and analyze the signals, the proposed system provides a stable sampling rate based on interrupts, ensuring a reliable signal for further analysis.

In order to reduce the number of sensors, it was introduced an adjusting factor to compensate the lack of that sensor. Referring to this factor, used to compute the CoP trajectory based on only two sensors per foot, it could be omitted or reduced if S4 and S5 sensors were replaced by sensors covering the whole metatarsal area. By doing that, the system could better estimate the total force, as it would detect greater pressure in the metatarsal area. It is important to mention that the CoP calculation is based in the total force exerted by the body to supporting area of the feet. So, the most the area measured, the best CoP estimation. It could be inferring that the values achieved using FSR are smaller compared to the force platforms values. This is expected due that the FSR based systems omit certain sensing areas of the foot, such as the region of the fingers, so that the force detected is lower and the values should be also lower.

However, it is important to note that, according to the presented results, it is not necessary to measure the whole pressure on the foot insole for the system to be capable to detect differences inter-subject for the typical Romberg's tests. Thus, this work demonstrate how is possible to assess the CoP using only two low cost and widely available commercial sensors.

As a result from above, the values obtained using the presented system could not be directly compared with valued obtained using force platforms, however, actually the indexes values of balance are quite variable among subjects, so there not exist a common range of values widely accepted by the community as the standard values (Koltermann *et al* 2019), instead, the balance tests are carried out testing the subject under different conditions and evaluating the difference. Thus, the proposed system allows, due to the portability and low cost, to be implemented and used in small clinics, rehabilitation offices or places were the patients usually develop their life, without the necessity to move to a specialized laboratories with force platforms or stabilometers.

The limitations of the proposed system are the following:

It is intended for static posturography by now, although in future research it could be explored their use for dynamic balance, modifying the system.

The position of the feet in the mediolateral axis is fixed, limiting by now the use of the system to subjects who can perform a typical Romberg´s test. However, the position of the sensors could be easily modified, for subjects with other characteristics, such as children or people with special needs.

The Table 6 compares the proposed system with WBB and commercials stabilometers systems.

Table 6. Comparation of the main characteristics of portable systems to measure the CoP.

	Proposed system	WBB®	Biosway®
Platform weight (g)	1170	3500	NA
Total system weight (g)	1830	6000***	20000
Platform Size (cm)	45x40x2	50x30x5	54x48x7
Price (USD)*	500	90*	8,000
Sampling rate (Hz)	100	~63	NA
Indexes	24	-	5
Battery autonomy	2 h	2 h**	ND

* Price only for the platform, but it requires a computer that could increase the cost in more than 600 USD, depending on the computer characteristics.
** Will depend on the battery of the computer, the battery for the WBB last months, depending on the use.
*** Considering a lap top weighting around 2.5 kg.
NA: Information not available
ND: The system does not have such characteristic.

It can be observed from Table 6 that, comparing the proposed device against WBB based systems and portable commercial stabilometers, it presents advantage in portability (weight, size), cost and number of indexes calculated. Is true that the cost estimated for the system presents is calculated at prototype level, which could be reduced if manufactured in mass.

Conclusions

In conclusion, a low cost, portable and standalone system was presented in this work. It was demonstrated that using only two sensors per foot it is possible to evaluate changes in balance due to different conditions as well as using three sensors. The system proposed is capable to calculate the CoP and 24 indexes commonly used in balance assessment using an embedded electronic device and to save the data for posterior analysis if required. Thus, the proposed system represents an alternative to evaluate the balance in environments outside laboratories at low cost.

The system is capable to detect differences in subjects due to past injuries, or age in preliminary test, although more research must be done to validate this.

This system could be applicable to evaluate changes in balance derived from ageing, neurological and musculoskeletal disorders, effects of drugs and lifestyle in balance, among others, as long as they can perform a typical Romberg´s test.

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