A real time study of the human equilibrium using an instrumented insole with 3 pressure sensors

Hussein Abou Ghaida, Serge Mottet and Jean-Marc Goujon

Abstract— **The present work deals with the study of the human equilibrium using an ambulatory e-health system. One of the point on which we focus is the fall risk, when losing equilibrium control. A specific postural learning model is presented, and an ambulatory instrumented insole is developed using 3 pressures sensors per foot, in order to determine the real-time displacement and the velocity of the centre of pressure (CoP). The increase of these parameters signals a loss of physiological sensation, usually of vision or of the inner ear. The results are compared to those obtained from classical more complex systems.**

I. Introduction

Our study concerns the human equilibrium in the scope of the ambulatory e-health systems. Several studies seek to detect and prevent the fall of patients. One of the effective diagnostics remains the measurement of the "stabilogram": the total center of pressure (CoPT) displacement in the standing posture and the velocity of displacement [1-2]. Hewson et al [1] present a statistical study based on the velocity of CoPT displacement, allowing classification of elderly as fallers or non-fallers. A lack of physiological sensation such as vision, somatic sensation or due to vestibular problem causes a significant increase in the CoPT displacement and its velocity [3-6], resulting in unbalance and increasing the risk of falling [7-8]. For most studies, the patient has to be still standing on a force plate.

On the other hand, plantar pressure measurements are used in the diagnostics of foot related problems such as plantar foot pain [9-11]. In the quiet standing position or during walking, the plantar pressure highlights the compression of the soft tissues (skin, fat, ligaments, and muscles) under the feet. The plantar pressure distribution can be measured by using insoles made of a pressure sensor matrix. F-SCAN® mobile System [12] with 954 sensors or Pedar® with 256 sensors [13] are the most popular instrumented insole systems used by podiatrists [14]. Those systems are mobile but not convenient for day use ambulatory applications.

A fully ambulatory system can be imagined, based on a sensor network with low power consumption and wireless transmission. The number of data to be processed and stored should then be small and adapted to a real time processing. The idea is to integrate the data treatment into an embedded system.

In this aim, we will show that in most cases, reducing the collected data to 3 local pressure measurements per foot is sufficient to obtain a real time stabilogram. The measurements are made by three sensors fixed under a thin insole.

Firstly, we describe the pressure sensor system. A specific learning method has been developed to identify the parameters of the transfer functions used in the numerical methods. The centre of pressure calculation method is then applied to obtain the stabilogram.

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II. Material

A. Sensors

The sensors used for the experiment are of Force Sensing Resistors type. These sensors are simple to use, low cost and flexible. The sensor voltage responses exhibit an exponential behavior (1):

$$V(F) = a_1 \cdot (1 - \exp(-a_2 \cdot F)) + a_3 \qquad (1)$$

The sensor voltage is amplified before analog to digital conversion. The amplified signal of each sensor may differ, due to the characteristic spread of the components and amplification electronics. The responses of the amplified sensors voltage have been calibrated by applying forces between 0 and 50 daN. The identification of the 3 parameters a_i is made for each sensor. A numerical filter is used to remove the high frequency noise due to the high resistance of the sensor, the cable length and the electronic circuit power supply.

The sensors are fixed under a commercially available leather shoe insole. The thickness is 2 mm including the black elastomer under layer, as shown in Figure 1 A. To fit the shoe size, the insoles can be reduced, as in the case of the F-SCAN insoles, Figure 1 C. The leather insole can be set on the FSCAN one, as shown in figure 1 B.

B. Learning method

We use the FSCAN system as a reference for pressure distribution under each foot.

The first step consists of measuring simultaneously the pressure distributions and the sensor responses for different postures. This learning step is performed to record the data necessary to identify the parameters of the transfer functions used in the numerical methods.

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The plantar pressure maps are recorded by using the F-SCAN® system (from Tekscan ®). The voltage delivered by the pressure sensors are recorded by using analog to digital converters from National Instrument interfaced by Lab View®. Both systems are externally trigged at the frequency of 20 Hz.

The posture has to vary during the recording in order to generate a data base for the numerical methods. This variation is natural during walking, but when in a standing position the subject is asked to exaggerate his posture in mediolateral and anteroposterior directions in order to get enough displacement amplitude to calibrate the sensors.

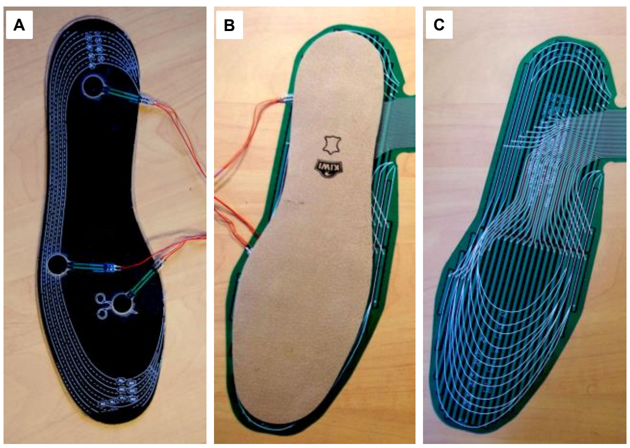

Figure 1. A) Bottom view of the leather insole with the sensors. B) Top view of the leather insole placed on a FSCAN insole. C) View of the FSCAN insole with its 950 sensors.

III. Method

A. Reference system of feet

The total centre of pressure, CoPT, derives from the feet plantar pressure repartition. The CoPT is a function of the feet position on the ground. In quiet standing posture, the coordinate origin on the ground can be defined as the mid-distance between the ankles. The x direction, oriented from the left ankle to the right one, corresponds to the mediolateral movements. The gap between the ankles is Δx . The y direction is perpendicular and corresponds to the anteroposterior movements (see Figure 2).

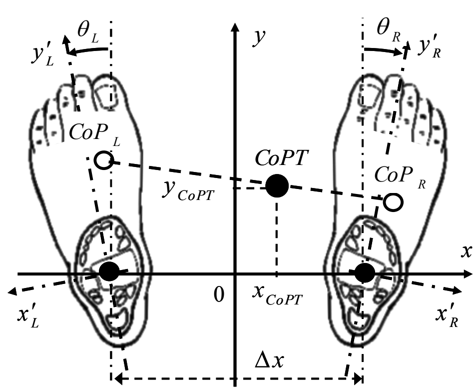

Figure 2. Foot system of reference

For each foot (L for left and R for right foot), the ankle is set as the origin of the internal coordinates. The $y'_{L,R}$ directions are given by the ankle and the second metatarsus of each foot (Figure 2).

The posture, depending on feet turned out or turned in, is defined by the angle $\theta_{L,R}$ made by the direction $(y'_{L,R})$ of each left and right foot with regard to the main (y) direction: $\theta_{L,R} > 0$ for turned out and $\theta_{L,R} < 0$ for turned in (Figure 2).

The total force applied to the foot is $F_{L,R}$, and the components of the projection of the force moment on the ground plane with regard to the ankle are $\breve{M}x'_{L,R}$ and $\breve{M}y'_{L,R}$.

B. Foot centre of pressure from FSCAN maps

By using F-SCAN system, the pressure data of $N = 950$ pixels per foot are available. For each foot, the data are the force f_i^k applied to each pixel $k = 1 \dots N$ with coordinates (x'_k, y'_k) at each time step $i = 1 \dots M$. The order of time step number is $M \approx 600$ for a 30 second recording.

At each step $i = 1 \dots M$, we proceed to the calculation of the total force applied to each foot (2):

$$F_i = \sum_{k=1}^{N} f_i^k \tag{2}$$

and the components of the moment along x' and y' (3):

$$\breve{M}x'_i = \sum_{k=1}^{N} x'_k \cdot f_i^k \text{ and } \breve{M}y'_i = \sum_{k=1}^{N} y'_k \cdot f_i^k \tag{3}$$

C. Foot centre of pressure from 3 discrete pressure sensors

The number of sensors per foot is $L = 3$. For each equilibrium time step i , the signal of the sensor j is fc_i^j . The main model hypothesis supposes 3 linear forms to describe the total force (4) and the moments from the sensors (5) and (6) which are:

$$\tilde{F}_i = \sum_{j=1}^{3} Cf_j \cdot fc_i^j \tag{4}$$

$$\tilde{\breve{M}}x_i = \sum_{j=1}^{3} Cmx_j \cdot fc_i^j \tag{5}$$

$$\tilde{\breve{M}}y_i = \sum_{j=1}^{3} Cmy_j \cdot fc_i^j \tag{6}$$

where Cf_j; Cmx_j; Cmy_j are 3 sets of 3 unknown coefficients to be determined for each foot. For each foot, the 9 coefficients are identified from the moments and the total force calculated from FSCAN measurements. This hypothesis will be verified through the results.

The coefficients Cf_j, Cmx_j, Cmy_j of the linear form (4, 5, 6) are calculated with the method of least squares applied to the three errors S_F , S_{mx} and S_{my} (7, 8, 9).

$$S_F = \sum_{i=1}^{M} \left(F_i - \sum_{j=1}^{L} Cf_j \cdot fc_i^j \right)^2 \tag{7}$$

$$S_{mx} = \sum_{i=1}^{M} \left(\breve{M}x_i - \sum_{j=1}^{L} Cmx_j \cdot fc_i^j \right)^2 \tag{8}$$

$$S_{my} = \sum_{i=1}^{M} \left(\breve{M}y_i - \sum_{j=1}^{L} Cmy_j \cdot fc_i^j \right)^2 \tag{9}$$

For each of the equation (7,8, 9), the calculation method leads to solve a Cramer system of dimension 3.

D. Total CoP calculation

The moment components $\breve{M}x'_{L,R}$, $\breve{M}y'_{L,R}$ and the total force $F_{L,R}$ determined from FSCAN or instrumented insoles allow the calculation of the total centre of pressure, CoPT, of the body. The calculation of the CoPT position from the plantar pressure distribution depends on the feet positions defined by the gap Δx between the feet and the feet angles $\theta_{L,R}$ which are measured beforehand by the podiatrist.

The x_{CoPT} and y_{CoPT} coordinates of the CoPT derive directly from the two following relations (10, 11):

$$x_{CoPT} = \frac{\breve{M}x'_L \cdot \cos\theta_L + \breve{M}x'_R \cos\theta_R - \breve{M}y'_L \cdot \sin\theta_L + \breve{M}y'_R \cdot \sin\theta_R}{F_R + F_L} + \frac{\Delta x}{2} \cdot \left(\frac{F_R - F_L}{F_R + F_L} \right) \tag{10}$$

$$y_{CoPT} = \frac{\breve{M}y'_L \cdot \cos\theta_L + \breve{M}y'_R \cdot \cos\theta_R + \breve{M}x'_L \cdot \sin\theta_L - \breve{M}x'_R \cdot \sin\theta_R}{F_R + F_L} \tag{11}$$

In a standing posture the feet position can be different from a patient to another. But in practice, the foot gap Δx is generally about 20-25 cm, which is approximately equal to the width of the hips.

Therefore, knowing the feet position, one can determine the CoPT displacement, and especially its velocity, from the plantar pressure distribution in a static standing posture.

The plantar pressure of eight male and female healthy subjects, have been measured for a time of 1 minute. The insoles were spaced at 22 cm, which corresponds approximately to the natural feet position. A normal feet position and a comfortable posture use slightly turned out feet with an angle less than 20°.

IV. RESULTS

Voluntary high amplitude anteroposterior and mediolateral displacements of the CoPT allow checking of the quality of the method. The movement is large; the force passes from foot to the other one in the mediolateral movement, and oscillates from toe to heel in the anteroposterior movement. These measurements were performed to identify the coefficients Cf_j; Cmx_j; Cmy_j for each sensor.

For a 70 kg subject, the Figures 3 A and 3 B show the x and y CoP displacements in function of time. Figure 4 C shows the stabilogram over a period of 30 seconds. The (blue) curve in the background is obtained by using the FSCAN system and the (yellow) curve in the foreground with 3 sensors per foot.

After calibration of the sensors, a test is made to measure the CoP in quiet standing posture. For a healthy individual, small displacements of the CoP are observed.

In order to demonstrate the interest of the measurement in the case of small oscillation, the experiment was performed on patients with closed eyes. The loss of visual sensation reduces the spatial reference and generates a slow drift mainly in the anteroposterior oscillation.

Figures 4 D shows the stabilogram while standing with closed eyes. The mediolateral displacement is small in the order of 3 cm. The good correlation between curves can be observed.

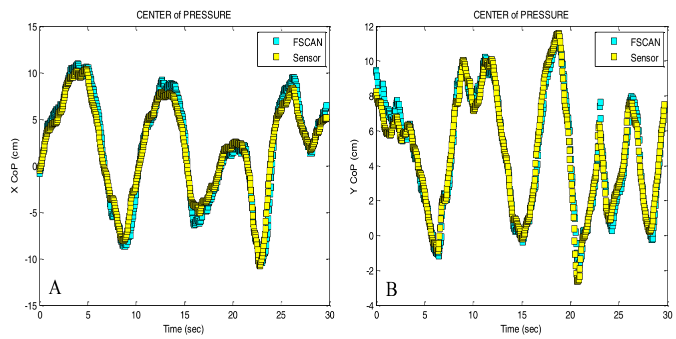

Figure 3. CoPT displacements versus time. A mediolateral and B anteroposterior.

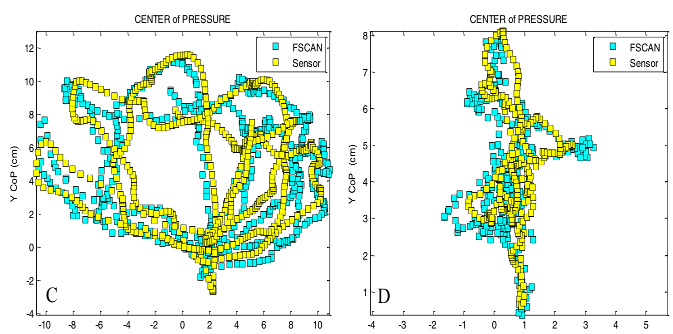

Figure 4. Stabilograms : C in forced movement, D in standing posture.

V. DISCUSSION

A set of 3 pressure sensors per foot allows the determination of the centre of pressure, CoPT, with real time processing and low power consumption. The sensors, located at the heel, the first and the fifth metatarsus, are fixed under a leather insole. The method uses the total force and moments applied on each foot to calculate the CoPT. Mathematically, the exact positions of the sensors do not have to be known. Moreover the method allows a total position shift of the foot with respect to the insole of $\pm 1\,cm$ in both x and y

directions, without significant effect on the result of the CoPT displacement. The insole being in a shoe means the possible shift is smaller than this.

Sets of 4 or 5 sensors per foot have been tested and the improvements made on the measurements of the CoPT are of the order of 0.5 mm along x and of 1 to 2 mm along y. However, with 3 pressure sensors per foot the accuracy on the CoPT in a standing posture is not affected.

The sensor coefficients are extracted from plantar pressures maps measured with F-SCAN system instrumented insoles. Two forces plates could be used to extract each foot sensor parameters (only one force plate for both feet is unusable) but it is not convenient. The results presented were obtained with a gap between the ankles Δx of 22 cm, and angles $\theta_{L,R}$ of 10° corresponding to a natural standing posture. But, in fact, the uncertainty on Δx does not affect significantly the measurement of the total CoPT.

The results show a good correlation between the CoPT measured with F-SCAN and with the 3 sensors insoles. The root mean square (RMS) error was calculated for the mediolateral and anteroposterior displacement. For the 9 experiments of 60 seconds each, the RMS error stays between 4.3 mm and 7.8 mm in mediolateral displacement, 2.4 mm and 5.9 mm in the anteroposterior. The RMS errors calculated for CoPT measured during standing posture varies of 3±2 mm in the mediolateral and 2±1.5 mm in the anteroposterior. In [15], the authors present an instrumented insole with 10 pressure sensors. They compare the CoP obtained from the instrumented insole with force platform measurements during a step. The RMS errors in the mediolateral and anteroposterior displacement range from 7 to 14 mm and 13 to 24 mm respectively.

Different patients, whose foot sizes vary between 37 and 46 (European size) have been tested without changing the position of the sensors on the insoles. The white lines in figure 1A show the different locations of the foot as function of the foot size. The forefoot sensors are always located under the first and fifth metatarsus for smaller size. Changing the size of the foot changes mainly the heel position with regard to the corresponding sensor. We show that a standard insole with fixed sensors position could then be used for different foot sizes thanks to the method. Size adjustment to the shoe is simply made by cutting along the corresponding size line.

VI. CONCLUSION

We have shown that insoles with 3 pressure sensors per foot allow the determination of the displacement of the CoP. The amount of data to be acquired is low, only 3 pressure values by foot. A sensor network with low power consumption can be considered. The digital transformation of these measurements into useful parameters, force and moments is simple and fast.

The sensors are to be used in a portable ambulatory monitoring system, with low power consumption and wireless transmission (Wi-Fi, Bluetooth…). The reduced number of data would be transmitted to podiatrist, or locally processed within a smart phone processor. This system could be suitable for balance studies of the elderly or for the analysis of walking and the activities of the daily living. These applications will be presented elsewhere. A future work will be made comparing healthy and unhealthy subjects.

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