Autonomous System For Movement Monitoring

Marc Torrent, Cecilio Angulo, Cristóbal Raya

Abstract—**Falls are a major source of concern for the health and quality of life in the elderly. This study presents a new prototype of wearable device based on accelerometers for long-term home monitoring. The principal aim of the device is to prevent falls by detecting balance disorders in the wearer. The objectives pursued in its design are the capability to work in uncontrolled environments and the wearer comfort and easiness. Validation of the device is done against a medical balance analysis instrument based on force plates (posturograph), and some of the results are presented. Finally there is a discussion on the objectives accomplished and possible future work in the project.**

I. INTRODUCTION

POPULATION in developed countries is ageing progressively and this is increasing the number of disabled people. In Spain, for example, there are 7.7 million of people older than 65 years. The 32.2 % of them presents some sort of disability, and this percentage is greater as the age range grows [1], [2].

Statistics confirm that the third part of the population older than 65 years suffers falls each year. This falls are one of the major causes of disability and dependency as they can cause major injuries. For instance, the 90% of hip fractures in the elderly are caused by a fall [3], [4]. This is a dramatic factor in disability and mortality as the 33 % of the old people that suffers from a hip fracture die in the next 12 months, and a 60 % develops some sort of disability that difficult their daily activities and makes their quality of life worse [5]-[8].

The early detection of fall risk persons is the most efficient way to not only prevent hip fractures, but also lower dependency, institutionalization, and mortality in the elderly. The existent methods for the falling risk estimation are based in interviews with the patient about his fall historical [9], or in an exploration of gait and equilibrium [10]. These methods have numerous disadvantages as they take on account only punctual determinations of the state of the patient. For this reason they can not register alterations caused by ambient factors or by circumstances that daily life involves.

Manuscript received April 30, 2008. This work has been partly supported by the Spanish MEC project ADA-EXODUS (DPI2006-15630-C02-01). First author is supported by an AGAUR research grant.

Marc Torrent is with the Technical Research Centre for Dependency Care and Autonomus Living (CETpD) of the Technical University of Catalonia (UPC), Vilanova I la Geltrú, PO 08800 Spain (phone: +34 93 896 72 78; e-mail: marc.torrent-ponch@ upc.edu).

Cecilio Angulo and Cristóbal Raya are with the Technical Research Centre for Dependency Care and Autonomus Living (CETpD) of the Technical University of Catalonia (UPC), Vilanova I la Geltrú, PO 08800 Spain (e-mail: {cecilio.angulo; cristobal.raya}@upc.edu).

The daily activity monitoring is a good source of information for a fall risk diagnosis because it can provide information on the habits, quality of movement, and unbalance historic for example. Inertial sensors such as accelerometers and gyroscopes have proven to be useful in the analysis of human motion [11] with the advantage that they can be integrated in some sort of wearable device. Accelerometers have been successfully used to classify daily life activities without supervision [12]. Also they have been used to automatically detect more specific events such as falls [13], [14]. In fall risk prediction accelerometers and gyroscopes have been successfully used to evaluate balance of a subject based on specific postural transitions such as sit-to-stand movements [15], [16], or specific exercises such as maintain balance with eyes closed on foam [17].

The aim of this paper is to study the use of a new prototype of wearable device for long term monitoring of gait and balance using accelerometers. First it is focused on the design of a device that can be used all day and ubiquitously (in the patient daily life) and be usable and as non invasive as possible for the patient. Secondly, experiments with different subjects were carried out in order to validate the device and to determine how many sensors are needed to take a measure of a subject's balance and what are the best locations.

II. METHODS

The followed methodology was based, first, on the design and building of a wearable device for kinematic parameters recording. Special attention was taken on making it usable in uncontrolled environments and comfortable to wear by the end user. The second step was

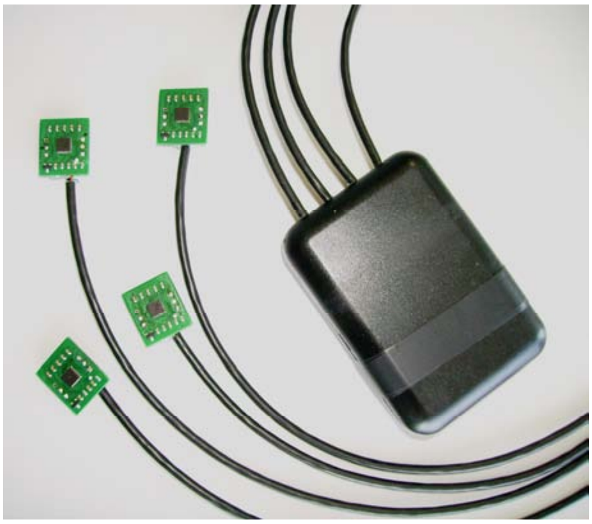

Fig. 1. Image of the wearable device with the central unit (IHU) and the four wired sensor probes.

to validate the prototype. It was done taking measures of the balance of different subjects in controlled exercises and comparing them with a medical posturography instrument based on force plates.

A. *The wearable device: IHU*

The wearable device is composed by a central Intelligent Hardware Unit (IHU) [18] and five sensor probes as shown in fig. 1. The IHU is the central part of the system, and have the function of sensor data handling. It features a microcontroller based process unit, a communications module, external memory, and interfacing circuitry. The core of the IHU is a dsPIC (Microchip Technology Inc.), a hybrid microcontroller with basic digital signal processor (DSP) features that enables the system with mathematic capabilities. The microcontroller operates at 20 MIPS (Mega Iterations Per Second) and have an internal program memory of 24Kbytes which allows a wide range of programming possibilities. The dsPIC uses a digital SPI (Serial Pheriperal Interface Bus) interface to communicate with the sensors in order to provide robustness to the readings. Another interesting feature of the dsPIC is his low power consumption along with its low power operation modes which enhances battery life.

The IHU can operate in two different modes: online mode or offline mode. In the online mode the sensor readings are transmitted in near real time via Bluetooth radio. The IHU integrates a Bluetooth transceiver Parani-ESD200 (Sena Technologies Inc.) capable to establish a wireless communication with another device using the RFCOMM protocol (virtual serial RS232 connection). This is a much extended protocol, so it enables the IHU to communicate with a large range of devices including laptops, handheld devices and mobile phones. The alternative is the offline mode, where the sensor readings together with a time stamp are stored in an external memory. The memory used is a microSD flash card of 1 or 2 Gbytes capacity. The main features of this memory are its reduced size, high capacity, low power consumption and the possibility to use a standard SPI interface to communicate with de microcontroller. Also, it gives easiness of use to the final user as it can be extracted and used directly on a computer to read the data stored.

The device, including the sensors, is powered by a single Lithium Polymer rechargeable battery with a capacity of 610 mAh. With this battery the system can run for about 6 hours without interruption, and then it needs to be recharged. All the electrical interface circuits including the battery charger, battery monitor, battery protection and voltage regulation are integrated in the own device. The size of the IHU with the battery and the casing included is 80x56x25 mm. and has an approximate weight of 30 gr.

B. *The wearable device: sensor probes*

The device is composed by five sensor probes wired to the central IHU (four wired plus one integrated on the IHU), to measure accelerometry in different parts of the body. These probes are much smaller than the IHU, they are composed by a tiny board of 17.8x20.3 mm. This design with tiny sensor probes connected to a central unit provides a more user friendly device. The probes can be attached to specific locations without being a major impediment for the user, and the central unit can be easily attached (to the belt for example). Furthermore, the use of probes can lead to a future integration of sensors in electronic textile and garment [19].

The sensor used is a MEMS 3-axis accelerometer H48C (Hitachi Metals Ltd.) with a measuring range of ±3 g and a sensitivity of 2.78 mg/mV. Each sensor is interfaced with the microcontroller using a 12bit analog-to-digital converter with SPI output. Due to the sampling, the sensitivity of the sensors is converted to 2.22 mg/LSB. The five sensors share the same SPI bus to communicate with the microcontroller and they are read sequentially at a rate of 45 samples per second. Prior to sending or storing the sensor data, the microcontroller applies a low-pass Finite Impulse Response (FIR) filter of 10th order and cut-off frequency of 2 Hz, in order to eliminate noise and unwanted high frequency signals.

C. *Device Validation*

Device validation was carried out using a Balance Master (Neurocom International Inc.) posture and equilibrium analyzer (posturography analyzer). The function principle of the posturograph is to measure the force exerted by the feet over a force plate. Using these measures the device is capable to calculate the centre of gravity (COG) of the patient and its frontal and lateral deviation in degrees over time.

Acceleration does not give a direct measure of COG inclination, although accelerometers can give the inclination respect to the gravity vector very precisely when measured in static. So in static activities (i.e. standing still) inclination of the body measured with accelerometers can give values very similar to the inclination of the COG of the body. The aim of the validation was to test the similarity of the results of the two devices in near static activities.

To calculate the inclination from the accelerometer readings, two reference frames can be fixed: the earth reference frame G_e, and the accelerometer reference frame G_a. The relation between two frames can be described by a rotation matrix as follows,

$$G_e = {}_a^eR \cdot G_a \tag{1}$$

where in equation (1) ${}_a^eR$ is the Euler angles rotation matrix with roll(φ), pitch(θ) and yaw (ψ), as seen in equation (2).

$${}_a^eR = \begin{bmatrix} c\theta \cdot c\psi & -c\phi \cdot s\psi + s\phi \cdot s\theta \cdot c\psi & s\phi \cdot s\psi + c\phi \cdot s\theta \cdot c\psi \\ c\theta \cdot s\psi & c\phi \cdot c\psi + s\phi \cdot s\theta \cdot s\psi & -s\phi \cdot c\psi + c\phi \cdot s\theta \cdot s\psi \\ -s\theta & s\phi \cdot c\theta & c\phi \cdot c\theta \end{bmatrix} \tag{2}$$

The gravity direction in the earth reference frame is given by,

$$G_e = \begin{bmatrix} 0 & 0 & -1 \end{bmatrix}^T \tag{3}$$

so from equations (1), (2) and (3) we obtain equation (4).

$$G_a = \begin{bmatrix} s\theta & -s\phi \cdot c\theta & -c\phi \cdot c\theta \end{bmatrix}^T \tag{4}$$

Therefore, operating equation (4) we can isolate angles φ and θ.

$$\theta = \text{atan2}\left(G_a(1), \sqrt{G_a(2)^2 + G_a(3)^2}\right) \tag{5}$$

$$\phi = \text{atan2}\left(-G_a(2) \cdot sign(c\theta), -G_a(3) \cdot sign(c\theta)\right) \tag{6}$$

In equation (5), θ is the pitch angle or the frontal inclination angle from the patient point of view, and φ in equation (6) is the roll angle or lateral inclination angle.

With these transformations the angles from accelerometry and posturography can be compared using correlation statistics to see if the results are the expected.

D. *Experiments*

The experiments were carried out with a sample of five volunteers with mean age of 29 years, mean height of 179 cm and mean weight of 88 kg. Each subject performed tree trials of four medic tests of balance analysis in the posturograph and wearing the accelerometry device. The first two tests carried out were part of the clinical test of sensory interaction on balance (CTSIB), which consists on standing still on top of a foam piece during ten seconds, first with eyes open and then with eyes closed. The third test was part of the unilateral stance (US) test that consists on standing still during ten seconds on a single foot and with the eyes closed. And the fourth test was the Tandem Walk (TW) test that consists on walking heel to toe from one end of the forceplate to the other. In total 20 trials were carried out, so the results can be considered significant due to the large amount of data captured.

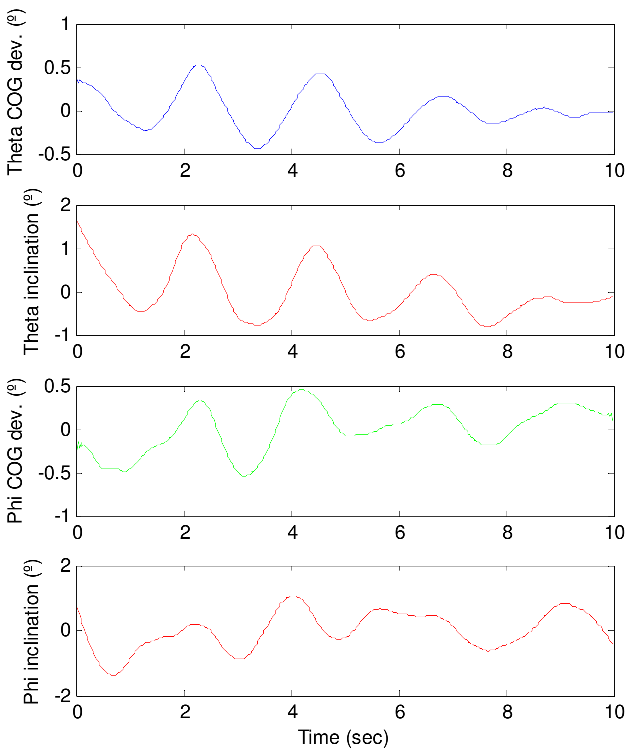

Fig. 2. Captured data from a CTSIB eyes closed test. Comparison with posturograph (a) and accelerometry (b) frontal inclination (theta), and posturograph (c) and accelerometry (d) lateral inclination (phi).

The accelerometry device was worn on the experiment subjects following a specific methodology. To attach the sensors firmly into the body and prevent relative movements and vibrations a special flexible straps were used. The sensors were placed one on each arm, sideways in the middle distance between the acromiale and the radiale. Another one on each tight, placed frontally in the middle distance between the trochanterion and the lateral epicondyle of femur. And one on the chest together with the IHU, placed frontally over the sternum.

The acceleration readings were captured online using the Bluetooth connection and a laptop. The device was synchronized in time with the captures from the posturograph using a synchronization signal.

III. RESULTS

To compare the results from the posturograph and the accelerometry device a Pearson linear correlation was computed. Since the data from the posturograph were highly filtered, another filter was applied subsequently to the accelerometry data using the Matlab software. The filter applied was a second order Savitzky-Golay smoothing filter with a window of 75 samples. Fig. 2 shows the similarities from the posturograph and the accelerometry device data during a CTSIB test.

The results obtained vary depending on the sensor location. For the compared parameters, the best correlations are obtained with the chest sensor followed by the tight sensors. It is not possible to make a direct comparison with the arm sensors readings and the COG inclination, so results are not showed. In each sensor results were compared for the frontal and lateral inclination angle and its change over time. Table I summarises the results obtained with the chest sensor when comparing the frontal inclination angle. The

Subjects			Correlations			
	Height (cm)	Weight (kg)	TW	US	CTSIB EO	CTSIB EC
1	180	76	0.80	0.87	0.75	0.87
2	181	87	0.72	0.86	0.83	0.85
3	190	114	0.73	0.82	0.82	0.92
4	171	70	0.70	0.77	0.76	0.85
5	174	94		0.78	0.76	0.83

Table I. Correlation coefficients of the frontal inclination angle in the different subjects and tests: Tandem Walk (TW), Unilateral Stance (US) and Clinical Test of Sensory Interaction on Balance with eyes open (CTSIB EO) and eyes closed (CTSIB EC).

correlations showed are the mean of the three trials for each test and each subject. The best correlation obtained was 0.96, and the mean of all test was 0.80. The comparison comprises 45 of the 60 tests carried out, The missing data cannot be used due to hardware problems during the realization of the tests. The best correlation achieved for the lateral inclination angle was 0.86, and the mean of all tests was 0.52.

IV. DISCUSSION AND CONCLUSIONS

The aim of this paper was to present a new prototype of wearable device for long-term home monitoring and fall risk evaluation of elderly people. To validate the device it was compared with the results from a posturograph measuring balance through the inclination of the COG across time. The experiments carried out in the laboratory gave good results when measuring the inclination with the chest sensor. The comparison of the frontal inclination angle gave a mean correlation of 0.80, which is a pretty good result having in mind that the compared measures were equivalent but not exactly the same magnitude. However, in the comparison of the lateral inclination angle, only a mean correlation of 0.52 was achieved. This error difference with the frontal inclination was not expected and can be attributed to an error in the measuring sensor. In any case, it will be a subject for further study in the future.

In general terms, the device has obtained pretty good measures of inclination and its variation over time, and this is good evaluator of the equilibrium of a subject [4]. This fact points out that the device could be used to get a balance appreciation of a subject who wears it. Further work in this aspect is to create a protocol and a set of conditions to define how to carry out the balance evaluation and how to interpret the results. To do this it is required to perform more experiments with end users (elder persons) that show different degrees in the Tinetti test of balance.

The design of the device with a central unit and a number of sensor probes responds to two future possible developments. One is the integration of the sensors in the wearer's clothes. With the use of sensor probes it is possible to integrate more easily the device and make it comfortable to wear since the probes are as big as a button. The only bigger part is the central unit, and this can be easily attached to the belt for example. The other future development is related to the number and location of sensors. The current prototype is composed by four sensor probes, but in future developments it is possible to add more or less sensors. It is known that the number of sensors affects directly the redundancy and reliability of the data, but there must be an agreement between redundancy and comfort.

The main reason to use sensors in various locations is that this device is intended for long-term monitoring in uncontrolled environments. In these conditions, prior to do a test of the balance one needs to know the current state and activity of the user. Various research groups are working on the development of daily activity classification devices with accelerometers [12], and they have achieved good results in the simulations. So one possible development is to first use a classifier to know what activity is the user carrying out, and then use the accelerometry data to make balance tests based on the activity. Future work in this project will be focused in achieve results in merging a activity classifier with the algorithms for balance evaluation.

ACKNOWLEDGMENT

The authors would like to thank the volunteers that took part in the tests carried out for this publication and specially Carlos Pérez, member of the CETpD-UPC research unit, for his technical collaboration during the development of this work.

REFERENCES

[1] *Las personas mayores en España. Informe 2004.* Madrid: IMSERSO, 2004.
[2] *Libro blanco de la dependencia.* Madrid: Ministerio de trabajo y asuntos sociales, 2005, Ref Type: Catalog.
[3] S. Boonen, J. Dequeker and W. Pelemans, *Risk factors for falls as a cause of hip fracture in the elderly*. Gent: Acta Clinica Belgica, 1993, 48(3): 190-4.
[4] M.E. Tinetti, *Clinical Practice, Preventing falls in elderly persons*. Waltham, MA: New England Journal of Medicine, January 2003, 348(1):42-9.
[5] A.L. Armstrong and W.A. Wallace, *The epidemiology of hip fractures and methods of prevention.* Brussels: Acta Orthopaedica Belgica, 1994, 60 Sup 1: 85-101.
[6] J.A. Stevens and S. Olson, *Reducing falls and resulting hip fractures among older women.* Atlanta, GA: MMWR Recommendations and Reports, March 2000, 49(RR-2):3-12.
[7] M. Cree, C.L. Soskolne, E. Belseck, J. Hornig, J.E. McElhaney, R. Brant et al., *Mortality and institutionalization following hip fracture.* New York: J. of the American Geriatrics Society, March 2000, 48(3):283-8.
[8] E.L. Hannan, J. Magaziner, J.J. Wang, E.A. Eastwood, S.B. silberzweig, M. Gilbert et al., *Mortality and locomotion 6 months after hospitalization for hip fracture: risk factors and risk-adjusted hospital outcomes*. Chicago, IL: J. of the American Medical Association, June 2001, 285(21): 2736-42.
[9] D.A. Ganz, T. Higashi and L.Z. Rubenstein. *Monitoring falls in cohort studies of community-dwelling older people: effect of the recall interval*. New York: J. of the American Geriatrics Society, December 2005, 53(12): 2190-4.
[10] S.G. Brauer, Y.R. Burns and P. Galley, *A prospective study of laboratory and clinical measures of postural stability to predict community-dwelling fallers*. Baltimore, MD: J. of Gerontology: Medical Sciences, August 2000, 55(8): 469-476.
[11] M. Mathie, A.C.F. Coster, N.H. Lovell and B.G. Celler, *Accelerometry: providing an integrated, practical method for long-term, ambulatory monitoring of human movement*. Physiol. Measure., Vol. 25, No. 2, R1-R20, April 2004.
[12] J. Pärkkä, M. Ermes, P. Korpipää, J. Mäntyjärvi, J. Peltola and I. Korhonen, *Activity classification using realistic data from wearable sensors.* IEEE Trans. on Information Technology in Biomedicine, vol. 10, No. 1, January 2006.
[13] U. Lindemann, A. Hock, M. Stuber, W. Keck and C. becker, *Evaluation of a fall detector based on accelerometers: a pilot study*. Med. Biol. Eng. Comput., 2005, 43, 548-551.
[14] N. Noury, A. Fleury, P. Rumeau, A.K. Bourke, G.Ó. Laighin, V. Rialle and J.E. Lundy, *Fall detection – Principles and Methods*. Proceedings of the 29th Ann. Int. Conf. of the IEEE EMBS, August 2007, pp. 1663-1666.
[15] B. Najafi, K. Aminian, F. Loew, Y. Blanc and P.A. Robert, *Measurement of stand-sit and sit-stand gyroscope and its application in fall risk evaluation in the elderly*. IEEE Trans. on Biomedical Engineering, Vol. 49, No. 8, August 2002.

[16] G. Constantini, M. Carota, G. Maccioni and D. Giansanti, *A new integrated kinematic sensor for the classification of sit-to-stand locomotion task*. 2nd Int. Workshop on Advances in Sensors and Interface, June 2007, pp. 1-5.

[17] D. Giansanti, *Investigation of fall-risk using a wearable device with accelerometers and rate gyroscopes.* Physiol. Meas., 2006, Vol. 27, No. 11, pp. 1081-1090.

[18] C. Raya, M. Torrent, J. Parera, C. Angulo, *Design of Intelligent Hardware Units for wireless online body monitoring.* Proc. on the 2nd Int. W. on Ubiquitous Computing and Ambient Intelligence, 2006.

[19] C. Mattmann, O. Amft, H. Harms, G. Tröster and F. Clemens, *Recognizing upper body postures using textile strain sensors*. 11th IEEE Int. Symp. on Wearable Computers, 2007, pp. 29-36.