

FALL DETECTION SYSTEMS

A Solution based on Low Cost Sensors

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Abstract: The problem of fall detection in elderly patients is particularly critical in persons who live alone or are alone most of the day. The use of information and communication technologies to facilitate their autonomy is a clear example of how technological advances can improve the quality of life of dependent people. This article presents a prototype developed with a low cost device (the gamepad of a known video console) using its Bluetooth communication capabilities and built-in accelerometer. The latter is much more sensitive than other similar devices integrated in mobile phones and much cheaper than industrial accelerometers. Besides its stand-alone use, the system can be connected to a generic remote monitoring system that has been developed as a software product line for use in aged people's residences.

1 INTRODUCTION

Dependence can, in general, be defined as the need for significant aid or assistance for the activities of daily life. Population aging is a factor that, in the future, will significantly increase the percentage of dependent population, due to the close relationship between dependence and age. The role of information and communication technologies (ICT) as a mechanism of social integration for older, disabled or dependent people in general is spreading rapidly among many sectors of the population. The increased costs of care and the geographic dispersion of an aging population favor the deployment of personalized services based on low cost distributed systems with ubiquitous computing tools. Wireless networks allow services adapted to different scales (large areas through wireless/cellular networks or home environments using short-range communication technologies such as Bluetooth) and provide overall support to these remote monitoring systems.

These technologies have generated huge expectations but we should ensure that their costs are affordable. We need to provide personalized care accessible to more people while reducing the costs of health systems. In addition to patients, other people with varying degrees of dependence can improve their level of autonomy: persons with different physical or mental disabilities, the elderly

who live alone or in residences, etc. The PATRAC project (in Spanish, "PATrimonio ACcesible", Accessible Heritage) is designed as a set of services that include monitoring of dependent visitors to cultural environments. In this context, one of the most common problems, especially in older people, is the detection of accidental falls, taking into account such facts as (Salva et al., 2004)

- 30% of people over 65 fall at least once a year.
- Fear, anxiety and depression are rising due to the risk of falls.
- Falls are responsible for 70% of fatal injuries for people over 75 years of age.
- A fall in an old man, even if it is a mild one, can cause irreversible damage or death.

For these reasons we determined the development of a system that automatically detects falls to augment the functionality of the abovementioned monitoring system. The advantages are clear, as these systems can increase the safety of older people, giving them the possibility of autonomy, while providing comfort to their families and caregivers. A secondary objective is to reduce the costs of caring for dependents. The original plan included:

- Finding or adapting a reliable fall detection algorithm that minimizes false positives and can detect all types of falls.

- Defining the technical characteristics of the sensor needed to implement a system using the selected algorithm.
- Implementing the algorithm on a prototype to check its performance in simulated falls and rapid movements (to eliminate false positives), leading finally to real situations.

The rest of the article details the proposed solution, beginning with the study of the detection algorithms published in the medical literature. As a result, an algorithm that combines the advantages of various methods is proposed and the requirements of the sensors needed for the implementation are stated. Section 3 shows how a low cost accelerometer can achieve those requirements and finally Section 4 presents the design and the results of the simulations carried out. Finally, similar products are compared, and the conclusions and future work close the paper.

2 FALL DETECTION

An initial review of the literature convinced us of the advantages of accelerometers as the most suitable type of sensors to detect falls. Although there are other alternatives, such as the use of gyroscopes (Bourke and Lyons, 2008), most works use two or three axes accelerometers (Bourke et al., 2004) (Chen et al., 2005). To design a reliable detection system based on these devices, the accelerations naturally present in the human body must be previously documented, both in normal movements and different types of falls. Various medical articles have studied these accelerations. When a person falls and hits the ground, his body suffers accelerations above those that occur when he is performing a normal activity. The work (Chen et al., 2005) studied the differences between sitting movements and falls by means of experiments with two two-axis accelerometers. Although the graphics were very similar, during a typical fall the acceleration is 7g, while the accelerations measured when a person sits down are less than 3g (about 2.6g where measured). Looking at the graphs presented in that article, it is noteworthy that, at the beginning of the fall, acceleration decreases (indicating the period of fall), but immediately there is a large peak indicating the impact against the ground (7g approx.). The accelerometer measurements, before and after the fall, are held at about 1g, as expected. Similar results, even with major peaks, were observed in lateral falls.

From the viewpoint of the type of falls, Lord et al. (Lord et al., 1993) found that 82% occurred when people were upright. The most common falls occurred while an elderly person is walking, slides and falls. Another study, conducted by (O'Neill et al., 1994), found that, of 180 crashes recorded, 160 were forward and, in 60% of these, the subject was taking a step forward with one bent knee and one foot in the air, the typical movement of a walking step.

With these studies as a reference, (Bourke et al., 2007) attempted to define the acceleration threshold that can automatically discriminate between normal body movements and different types of falls. The values of the accelerations were derived from daily activities performed by elderly people and simulated falls performed by young people. The first experiment involved ten elderly people, aged between 70 and 83, with a tri-axial accelerometer, placed first on the trunk and then on the thigh. The activities were sitting and rising from an armchair or a kitchen chair, walking 10 m, etc. The second experiment used ten young people aged between 21 and 29 who simulated six different types of falls. Although forward falls are more frequent, they also simulated lateral falls, as these often produce a great impact on the trunk and often result in fractures when they happen. The authors selected the lowest value of the accelerations recorded during simulated falls (upper fall threshold), and the largest of the smaller peaks (lower fall threshold). The smallest accelerations during a fall were about 3.5g but others were much greater, while normal activities usually produced accelerations of 1 to 2.5g, although sometimes there are activities, such as running or sitting, which can surpass this. In conclusion, the threshold of normal movements should be between 0.41g and 3.52g. The acceleration values outside this range could be considered potential falls. The success percentage of the algorithm, including false positives, was calculated with the accelerometer placed on the trunk and on the thigh. The best results were obtained for the trunk, with more than 90% correct hits. But false positives (false alarms) remained the real problem.

(Chen et al., 2005) used a different approach that took into account the unexpected changes in body orientation. They also studied the situations of repeated impacts to determine certain types of falls (on staircases, for example) that may be especially dangerous. Based on these studies, we propose an experiment using an algorithm that combines the orientation changes postulated by Chen and the

thresholds measured by Bourke. To carry out these measurements, the required sensor must have the following specifications:

- The accelerometer must be tri-axial.
- It must be capable of detecting accelerations over 3.52g and under 0.41g, as these are the fall thresholds. This requirement eliminates many of the available accelerometers, particularly those embedded in current mobile phones with no more than 2g sensibility.
- It must operate in a wireless environment.
- It must work for several hours. The battery life is the key point here.
- Its weight and size should be reduced, since it will be placed on the patient's body (preferably integrated in the patient's clothes).

3 THE WIIMOTE AS ACCELEROMETER

The Wiimote is the main controller from the popular Nintendo Wii game console. Its main features are the ability to detect motion in space and that of pointing to objects on the screen. The design of the Wiimote is not based on traditional video game controllers but is intended to be used with one hand in an intuitive way. Because of its low cost and its potential, there are many initiatives that are evaluating their possibilities. WiiHome is an application developed by (Lee, 2009) to control the home through home automation devices (you can turn on and off a light, the TV, an alarm, etc.). The CEDETEL Research Centre (CEDETEL, 2009) is developing a series of applications for rehabilitation and increasing cognitive abilities for disabled people. They use the Wiimote as a device that allows the movements of patients to be captured and recorded so as to monitor their assigned exercises and to automatically control the degree of personal improvement.

The gamepad detects the acceleration measured along three axes using a built-in accelerometer (Figure 1). The batteries can power the Wiimote for 60 hours using only the accelerometer function (very interesting for our requirements). It uses Bluetooth to communicate with the console but is detected by other Bluetooth devices such as PCs or mobile phones.

The built-in tri-axial accelerometer provides instant acceleration values. The maximum value that can be measured is about 7g, which complies with the requirements specifications. In repose, this

acceleration is 1 g, upwards. While falling, the Wiimote indicates lower accelerations, close to 0g. Once the arbitrary accelerations provided by the Wiimote are captured and assuming that the accelerometer response is approximately linear, we can use standard positions to calibrate the controller on a flat surface: two horizontal positions that provide the values (x_1, y_1, z_1) and (x_3, y_3, z_3) and a third upright position giving the vector (x_2, y_2, z_2) . Because the accelerometer records the force of gravity, the data received in the three positions should be matched with three orthogonal acceleration vectors, so that in each of the three positions indicated, two of the three components of each vector will be zero and the third 1 g. Using the values provided in real time by the command $(XValue, YValue, ZValue)$, see Figure 1), we can convert these values into three orthogonal vectors with respect to g. Given $x_0 = (x_1 + x_2) / 2$, $y_0 = (y_1 + y_3) / 2$, and $z_0 = (z_2 + z_3) / 2$ we have:

$$X = (XValue - x_0) / (x_3 - x_0)$$

$$Y = (YValue - y_0) / (y_2 - y_0)$$

$$Z = (ZValue - z_0) / (z_1 - z_0)$$



Figure 1: Wiimote and its accelerometer axes.

This transformation of the Wiimote raw data in position 1, as described previously, gives the values $(1,0,0)$, in the second position $(0,0,1)$, and in the third one $(0,1,0)$. Once calibrated, we have reproduced the studies of previous works with the Wiimote accelerometer, checking that the values of the accelerations measured for the same activities as the previous research give close enough values. In our case, the tests were carried out by three people aged between 23 and 50. These tests were walking, sitting, getting up (Table 1). The upper fall threshold (UFT) in the walking test using the Wiimote was 2.04, while in the study of (Bourke and Lyons, 2008), it was 1.99g. In the same test, the lower fall threshold (LFT) was 0.66g while in the cited study it was 0.62g, both close enough. The WiimoteLib library was used (Brian, 2009) to manage the Bluetooth connection.

Table 1: Acceleration values (measured with a Wiimote).

Walking	UFT	LFT	UFT	LFT	UFT	LFT
Person 1	1.70	0.60	2.04	0.57	1.76	0.65
Person 2	1.80	0.61	1.65	0.66	1.69	0.63
Person 3	1.72	0.62	1.74	0.58	1.93	0.54
Sitting	UFT	LFT	UFT	LFT	UFT	LFT
Person 1	2.63	0.65	1.60	0.73	2.00	0.57
Person 2	1.36	0.83	1.33	0.80	1.36	0.79
Person 3	1.48	0.59	1.49	0.72	1.36	0.72

4 DESIGN OF THE FALL DETECTION SYSTEM

The system was developed in response to a set of basic requirements, validated by the medical staff of a senior citizens' residence to help us in developing monitoring systems. The most representative are:

- The system must notify the medical staff in real time that the patient has suffered a possible fall.
- The system will send an alarm to the central server when a possible fall is detected.
- The system should allow the patient to know when an alarm has been sent to the server.
- The system must allow the patients to deactivate the alarm if they see they need no attention.
- The system should allow the patients to make an emergency call when they think they need medical attention.
- The central system should provide a status picture of all the connected sensors, including the battery status, at any time.
- The central system must associate a Wiimote with each patient to identify the received data.

NET and C # were used to develop the system, due to the ease of integration with the available platform libraries. The main actors are the patient and supervising personnel (usually medical staff but can also be a member of the family). An actor models the automatic data that are obtained every few seconds (the time interval is configurable). The current version has been developed for home scenarios (or in a small residence), since the limits are defined by the Bluetooth connection range (a maximum theoretical distance of 100 meters).

The overall system architecture is shown in Figure

2. The acceleration data are collected by an auxiliary computer, located at the home of the monitored person and analyzed in real time. If a situation reflects a possible fall, the vibration of the patient's own Wiimote indicates the problem and after a few seconds an alarm is generated to be sent to a central system via http using a generic Web service. This system allows alarms to be collected and data to be continuously monitored, including the patient location obtained from devices with built-in GPS (Laguna et al., 2009). The elapsed time from the moment of detection until the alarm is sent allows the person to cancel the alarm if it is a false positive.

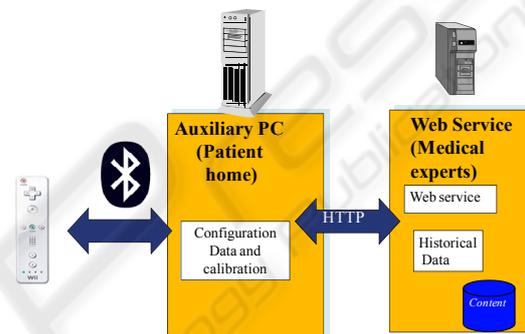


Figure 2: Architecture of the fall detection system.

The tests reflected in Table 2 were planned to check the effectiveness of the system. They were divided into two groups: normal movements (to detect false positives) and fall simulations by young people, due to the high risk that real falls represent for elderly people. For this purpose, the Wiimote was fixed to the hip of the subjects (Figure 3) and the system was installed on a standard PC, with a set of windows that continuously display the status of each registered sensor.

The tests have been encouraging. In total, 65 tests simulating falls have been conducted and the system has identified 55 possible falls with a success rate of 84.6%. One might consider this success rate to be low, but we must keep in mind the fact that the falls the system has not been able to recognize were all of the same type, a fall type resulting in the trunk remaining straight after the fall (“fall to a sitting position”). The way the detection algorithm is designed, based on shifting and impact, does not generate an alarm, as the sensor continues in a vertical orientation (like the trunk of the person). We are working on improving the algorithm, although one might think that if a person falls into this position (perhaps the least dangerous of the considered types of falls), there are

many chances that the patient can press the emergency button, also programmed in the sensor.

Table 2: Results of the fall detection tests.

Normal movements	Tests	False positives
Walking 20 meters	10	0
Going downstairs	10	0
Sitting on an arm chair	10	0
Going up 7 steps	10	0
Lying down in bed	10	5
Getting out of bed	10	5
Running 20 meters	10	0
Simulated Falls	Tests	Recognized falls
Front fall	10	10
Reverse fall	10	10
Lateral fall	10	10
Right trunk fall	10	0
Random fall	25	25

Concerning normal movement tests, the success rate was 88.8%. The system indicated 10 false positives of the 90 tests performed. In this case, false positives occurred in a specific type of movement, lying down and getting out of bed when the move was made without first sitting ("jumping from the bed"). In the tests where the subject first sat on the bed and then lay down (or got out) there were no false positives. Given that older people have limited mobility, it is rare they get out of bed quickly, so the problem should be minor in practical situations.



Figure 3: Correct placement of the sensor in the trunk.

Besides improving the algorithm, we are working on a second version for outdoor patient monitoring, using a Smartphone. The same algorithm has been implemented using the accelerometers present in

two types of mobile devices based on Windows Mobile (HTC Diamond and Omnia Samsung), joining the sensor and the PC functionality in a single device. The alarm can be sent via 3G Wi-Fi (or SMS to a configurable phone number). However, the results are not as reliable due to the lower range of accelerations measured.

5 RELATED WORK

Given the interest in the topic, many works have dealt with the development of devices and the associated algorithms to detect falls. (Degen et al., 2003) created "Speedy", a fall detector that operates via an accelerometer placed in a wristwatch. The algorithm uses a multi-stage approach: during the first stage, it looks for a high acceleration toward the ground, followed by an impact. Once the impact is detected, if a period of inactivity greater than 40 seconds follows, an alarm is activated. This device was successful in not producing false alarms, but was a disaster in falls other than front variants. It was unable to detect other fall types such as lateral or backward falls.

The Tunstall falls detector is a commercial system, developed by (Doughty et al., 2000), which uses a fall detection algorithm with two steps. They use two sensors: the first one detects the impacts, while the second considers the orientation. In short, when an impact is detected, the orientation of the system during the periods previous and posterior to the impact are analyzed and if there is a change of orientation the alarm is activated.

The obvious advantage of our system compared to existing products is its cost (less than 70 € including the cost of the sensor and a Bluetooth device that can turn any domestic PC into an alarm detection system).

6 CONCLUSIONS

This article describes a monitoring system based on an algorithm capable of detecting a wide range of falls and of eliminating many false positives. Based on published studies, the results have been reproduced in a satisfactory manner, improving them in some cases. Its performance has been tested in simulated falls and normal but relatively violent movements. The identified technical characteristics limit the useful sensors (tri-axial accelerometer with sensitivity better than 3.5g, wireless, light ...). A

basic architecture has been implemented, using a conventional PC connected via Bluetooth with a low cost device (the gamepad of a video game console).

Work in progress is devoted to integrating the system into a generic product line of mobile monitoring. Finally, to check the device in real patients, some tests have been scheduled in two residences that have shown interest in the device and have previously collaborated on the development of the generic application of continuous monitoring of physiological parameters.

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