THE DEVELOPMENT OF AN ACCELEROMETER AND GYROSCOPE BASED SENSOR TO DISTINGUISH BETWEEN ACTIVITIES OF DAILY LIVING AND FALL-EVENTS

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Abstract: This paper describes the development of an accurate, accelerometer and gyroscope based fall-event detection system that distinguishes between Activities of Daily Living (ADL) and fall-events. Using simulated fall-events onto crash mats (under supervised conditions) and ADL performed by elderly subjects, distinguishing between falls and ADL is achieved using accelerometer and gyroscope-based sensors, mounted on the trunk and thigh of the person. Data analysis was performed using MATLAB® to determine the peak accelerations and angular velocities recorded during eight different types of falls. A fall detection algorithm was proposed using thresholding techniques. Results from an evaluation of the detection algorithm show that a fall-event can be distinguished from an ADL with 100% accuracy using a single threshold applied to the resultant vector acceleration signal from a tri-axial accelerometer located at the chest. Thresholding was thus demonstrated to be capable of discriminating between an ADL and a fall-event, when those falls were simulated falls.

Keywords – accelerometer, gyroscope, fall detection

Introduction

Falls in the elderly are a major problem for today's society. Based on statistics from the United States, the following observations can be made: One of every three adults 65 years old or older, falls each year [1-3]. Falls are the leading cause of injury deaths among people 65 years and older [4]. In 1998, about 9,600 people over the age of 65 died from fall-related injuries [5]. Of all fall deaths, more than 60% involve people who are 75 years or older [4].

Some of the many consequences of a fall include fracture, significant injury and hospital admission [3] Furthermore, fear of recurrence can result in loss of confidence and curtailment of domestic and social activities culminating in a loss of independence and institutionalization [6]. With improved life expectancy the population of older adults is expected to increase dramatically over the next decade putting further pressure on the healthcare sector to increase resource allocation for the treatment of age-related accidents and illnesses.

Automatic detection of a fall would help combat injury severity resulting from a fall, by reducing the time between fall on-set and the arrival of medical attention. The existing common alarm system, the push-button pendant, is not satisfactory, as during a loss of consciousness or a faint the pendant will not be activated.

A number of different approaches to fall detection, have appeared in recent years [7]. These can mainly be separated into two groups, primary fall detection systems, which have the principal objective of instantaneously detecting falls, and only falls, and secondary fall detection systems, which indirectly detect falls by the absence of normal activities [7]. Both systems can be implemented as either worn devices or using embedded sensors in home [7]. This paper describes the development of a primary fall-detection worn system.

The end of a fall may be characterised by an impact and by near horizontal orientation of the faller following the fall and thus fall detectors must detect one if not both of these characteristics [7]. Most primary fall-detection systems detect the shock received by the body upon impact to determine if a fall has occurred and the typical sensor used is the accelerometer [8-11]. When attached to the body, accelerometers can be used to measure retardation of the body when it is arrested by the ground, following a fall.

Doughty et al [8] used accelerometers placed at four different locations (trunk, thigh, waist and wrist) of a jointed wooden mannequin to determine the thresholds associated with a fall. The mannequin performed five fall types altogether, knee-rigid falls in all three directions and two forward falls, both with knee-flexion, one involved the mannequin falling down a stairs. The results of this study concluded that the optimum sensor location was the trunk, between the chest and waist. However, no data is available, in the literature, on the performance of the system during simulated falls on human subjects or on the fall-detection accuracy of the system.

Noury et al [11] also used accelerometers to detect falls, but instead of detecting impact, the detection of rapid changes in posture were used to indicate that a fall had occurred. The system was tested using forward and backward falls but no lateral falls were performed in testing. An important feature of Noury et al.'s study was that the fall detection system was also tested for mis-detection of Activities of Daily Living (ADL) as falls. During this analysis, Noury et al. tested the system with a group of young subjects who performed a comprehensive series of activities such as; sit-to-lye on a bed, walking then bending down and kneeling, walking then laterally hitting a wall and sitting on a chair. Noury et al obtained a fall detection accuracy of 83% and an ADL detection accuracy of 79%. Thus 21% of ADL would be mis-detected as fall but more crucially, 17% of falls were not detected, which means that a potentially serious injury could occur to an elderly person and would go undetected with Noury et al's system, which is problematic.

Hwang et al. [9] used a tri-axial accelerometer and a gyroscope, both placed at the chest to successfully distinguish between falls and ADL, the system produced a fall-detection accuracy of 95.5% and no ADL were mis-detected as falls. The system was tested using only three young healthy subjects. Although simulated falls in all directions were performed, only two ADL activities were used to test the system which was performed once by each subject. There was 0% mis-detection of ADL as falls. Lateral falls were also occasionally not detected as falls, stemming from a lack of a gyroscope sensor in the lateral direction, acknowledged by the author.

To date fall-detection systems have used young subjects to test the ADL accuracy of their system. No study has used elderly subjects to perform this task. Elderly people may move differently than younger people as they may have less control over the speed of their body movements due to reduced body strength with old age, therefore, they may fall into a chair when sitting down instead of sitting in a controlled manner. In addition, since a fall detection device's target audience is elderly people, testing should account for the dynamics of the movements of these people. Thus, testing the extent of mis-detection of ADL as falls in a fall-detection system should involve elderly subjects.

This paper describes the development of an accelerometer and gyroscope based sensor capable of distinguishing between fall-events and activities of daily living (ADL). The eventual aim of the system is the monitoring of older adults while in their own home and in the event of a fall, alerting the emergency services to enable a quick response, thus, reducing the consequences of a fall. Accelerometer and gyroscope signals were acquired during simulated falls performed onto crash mats by healthy young subjects, which were compared with signals from Activities of Daily Living (ADL), performed by elderly people in their own homes. It is hypothesized that recruiting elderly people to perform ADL testing of a fall-detection system increases the robustness of the test methodology.

Materials and Methods

To develop the algorithm to distinguish between fall-events and ADL, two separate studies were performed:

1) A simulated fall-event study - used to establish a number of different thresholds that would indicate that a fall had occurred

2) An Activities of Daily Living (ADL) study – to determine the extent of miss-detection of ADL as fall events.

This study involved young healthy males performing simulated falls onto large foam crash mats under the supervision of a physical education professional. Longitudinal, sagittal and medial-lateral accelerations as well as pitch and roll angular velocities were recorded from the trunk and thigh during each simulated fall-event, figure 1. A total of eight different fall types were completed with each fall-type being repeated three times, by each subject.

The subjects were young (<30 years) healthy males. A total of 10 subjects were recruited for the study. The mean±standard deviation age, height and mass of the subjects were 23.7±2.2years, 1.78±0.058m and 75.9±5.1kg respectively. The exclusion criteria for this group was a history of any balance impairment, unexplainable spontaneous falls, neurological disease or uncorrected visual shortfall and all claimed to exercise regularly (>4 hours/week). All subjects, from this simulated fall-event study, gave written informed consent and the University of Limerick Research Ethics Committee approved the protocol.

The fall types were selected in order to best simulate the type of fall that may occur and cause injury to an elderly person. Thus, each fall was performed with the subject initially in a standing position. All the falls were performed onto large crash mats with a combined thickness of 0.76m.

The simulated fall strategies performed were:

Forward Fall: A forward fall occurs when a persons' centre of gravity has fallen outside their base-of-support (the area between and including the soles of their feet) and a recovery step or manoeuvre is not performed. The person will experience a loss of balance (LOB) as they are in an unstable state of equilibrium and they will descend to a horizontal position, maintaining an upright posture through the flight of the fall. The fall will finish with person coming to rest at a lower level, faced downwards.

The 2-stage Forward Fall: A two-stage forward fall occurs when a person first falls-to, or bends-at, the knees, thus commencing the fall-event, their upper extremity will then pivot about the knees and

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they will finish, lying in a horizontal position, faced down. This fall type was used to simulate fainting (or syncope), which is a common cause of failing.

The Backwards Fall: A backwards fall occurs when a person's centre of gravity has fallen out side their base-of-support, behind their feet. This type of fall will finish when the person comes to rest in a horizontal position, faced upwards with their back to the ground. An upright posture is maintained through out the coursed of this fall type.

The 2-stage Backwards Fall: A two-stage backward fall occurs when a person first bends-at the knees and then falls backwards to the ground. This type of fall will finish when the person comes to rest, face up, with their trunk in a horizontal position.

Lateral falls, left and right: A lateral fall occurs when a person's centre of gravity has fallen outside their base to the left or right of their feet. This type of fall will finish when the person comes to rest in a horizontal position.

The 2-stage Lateral fall, left and right: A two-stage lateral fall occurs when a person first bends-at the knees and then, laterally, falls to the ground. Impact occurs first to the knee and hip followed by impact to the arm and shoulder. Again, this type of fall will finish when the person comes to rest at a lower level on their side.

Subjects were advised to fall as naturally as possible within the guidelines provided and to initiate the fall with a slight movement of the body in the direction of their fall. A physical education professional supervised the subjects performing the simulated falls. All the simulated fall-events were performed in the sports building of the University of Limerick.

The second of two studies performed involves elderly people performing Activities of Daily Living in their own familiar environment while the same acceleration and angular velocity reading of the fall event study, were taken from the trunk and thigh, figure 1.

For the ADL study ten community-dwelling elderly, three female and seven male, were monitored while performing normal activities of daily living. The subjects ranged in age from 70 to 83 years old. (77.2±4.26yrs). Exclusion criteria for this study were; cognitive disorders, which would limit the comprehension or execution of the study, an inability to stand up from a seated or lying position without help, regular use of a walking aid including a cane, recent surgery (<1 year), or existence of an unstable medical condition. All subjects, from this ADL study, gave written informed consent and the University of Limerick Research Ethics Committee approved the measurement protocol.

In this study all ten elderly subjects performed a series of eight pre-selected ADL. Each ADL was performed three times by each subject. These ADL include:

- Sitting down and stand up from an armchair
- Sitting down and stand up from a kitchen chair
- Sitting down and stand up from a toilet seat
- Sitting down and stand up from a low stool
- Getting in and out of a car (drivers side)
- Sitting down on and getting up from a bed
- Lying down and getting up from a bed
- Walking 10m.

Each ADL started and ended with the subject in a standing position. All subjects performed each ADL three times, except for the walking activity, which was recorded for a distance of 10m. Each subject performed the relevant ADL using furniture that already existed in his or her own individual environment. All subjects were informed that they were not participating in a fitness test or illness diagnosis examination and to perform the activities as naturally as possible, considering the circumstances, and to take as much or as little time as they needed.

In both studies, trunk and thigh longitudinal, sagittal and medial-lateral accelerations and trunk and thigh pitch and roll angular velocities of the subjects were recorded using ADXL210 accelerometer and ADSRS300 gyroscope based sensors, respectively.

A portable battery-powered data-logger (Biomedical Monitoring BM42[1]) and associated computer-interface hardware, was used for data acquisition. For the simulated fall study, a special constructed platform was designed to facilitate falling onto crash mats safely.

The crash mats used were gymnasium mats whose combined thickness was 0.76m, designed so that fall events could occur without injury. The subjects stood on the wooden support platform, designed to safely support the weight of a fully-grown person. The dimensions of the platform were 1.22m (length) X 0.91m (width) X 0.76m (height). The platform and crash mats were level, which provided a safe level environment for the subjects to fall upon.

Two sets of sensors were used, a tri-axial accelerometer and a bi-axial gyroscope on both the trunk and thigh.

The accelerometer sensors consisted of two Analog Devices ADXL210 Accelerometers mounted perpendicularly to each other, with one accelerometer IC vertically mounted and the remainder horizontally mounted to achieve a tri-axial accelerometer sensor. This accelerometer arrangement thus measured longitudinal and tangential posterior/anterior and lateral, static and dynamic, acceleration of both the trunk and thigh body segments, figure 1.

[1] Biomedical Monitoring Ltd, Glasgow, Scotland.

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The second sensor set, a bi-axial gyroscope, consisted of two Analog Devices ADXRS300 gyroscopes. Both gyroscope-ICs were vertically mounted but perpendicularly orientated to each other, thus allowing the detection of angular velocity of the body, in both the sagittal and frontal planes described as pitch and roll angular velocity, figure 1. Only pitch and roll angular velocity were obtained, angular velocity about the yaw axis only served to measure the angular velocity of the subjects' rotation about the bodies' vertical axis, which was deemed unnecessary, as it would not have provided any extra valuable information about a fall.

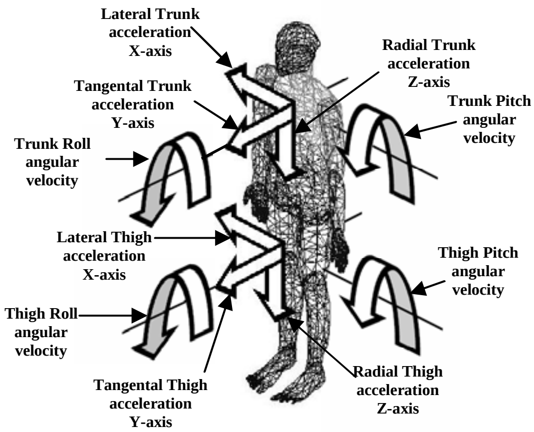


Figure 1: Accelerometer and Gyroscope axes orientation on the trunk and thigh.

The ADXL210 accelerometer has a measurement range of ±10g and outputs a voltage whose amplitude is directly proportional to the acceleration experienced.

The Analog Devices, ADXRS300 is a single axis angular rate sensor (gyroscope) capable of measuring angular velocities in the range $\pm 300^{o}$/s and provides an analogue output voltage.

Part of the requirements for the arrangement of accelerometers and gyroscopes adopted for this study was that a subset of this sensor arrangement (vertical accelerometer on trunk and thigh) also provides subject mobility data (amount of time spent sitting, standing, lying and walking) [12]. It was considered that the ideal arrangement would be that the sensor configuration for mobility assessment would also be sufficient for fall detection.

For both the ADL study and the simulated fall-event study, the participants were fitted with the ADXL210 accelerometer and ADXRS300 gyroscope sensor as discussed. These sensors were held in place using Velcro straps. The investigator held the data-logger while each activity was performed, to avoid damage to the device.

At the end of each recorded activity, the data-logger data was downloaded to a computer using an USB memory card reader for later analysis using MATLAB®

The following data analysis flow was performed:

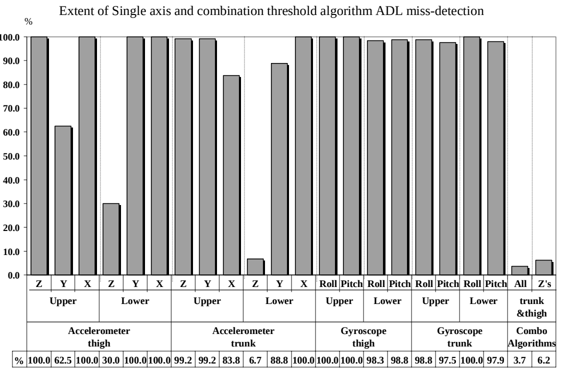
1) Conversion of tri-axial Accelerometer and bi-axial Gyroscope sensor voltages to acceleration and angular velocity using bench calibration data.
2) Each signal was low-pass filtered using a second-order low-pass Butterworth 2-pass digital filter, with a cut-off frequency of 20 Hz.
3) A resultant signal was then produced for both the tri-axial accelerometer and the bi-axial gyroscope, using the formula $\sqrt{X^2+Y^2+Z^2}$, for the tri-axial accelerometer and $\sqrt{X^2+Y^2}$, in the case of the bi-axial gyroscope, these new signal are referred to as resultant vector signals.
4) Maximum, positive and negative peak acceleration and peak angular velocity readings for all separate axes and the resultant signals were noted, during each Fall-event and each ADL activity.
5) For each axis, of each sensor, on both the trunk and thigh sites, the smallest positive going and negative going peak value from all falls completed were determined, thus providing an Upper and Lower threshold for each axis of each sensor. This threshold corresponds to the least severe fall-event experienced for that sensor axis during the course of 240 falls in total.
6) Each Upper and Lower peak acceleration and angular velocity threshold readings were then compared with each of the corresponding signals for the 240 ADL events recorded to determine the extent of mis-detection of ADL as fall-events for each individual sensor axis.
7) Each Upper and Lower threshold for the resultant acceleration and for the resultant angular velocity were then used to test the resultant signals for each of the 240 ADL events to determine the extent of mis-detection of ADL as fall-events for each resultant signal.
8) The extent of mis-detection of ADL as fall-events when the Upper and Lower thresholds for each axis and for the resultant were utilised for both sensor sets. In addition, a number of different thresholds were "ANDed" together and a composite detection algorithm constructed. Using the new threshold approach the extent of mis-detection of ADL as fall-events was determined.

Results

The longitudinal-axis, Lower threshold, from the trunk and thigh accelerometer sensor, was by far the most successful thresholds at distinguishing between fall-events and ADL, Table 1. With the longitudinal-axis trunk and thigh, lower thresholds mis-detecting 6.7% and 30% of ADL as fall-events, respectively The 6.7% of ADL mis-detected as fall-events by the longitudinal-axis trunk, lower threshold, was mainly attributed to one mis-detected ADL, "lying on a bed", which was responsible for 15 of the 16 mis-detected ADL, "sitting on an armchair" was the other offender. The longitudinal-axis thigh, lower threshold, fall mis-detection percentage (30%) was mainly attributed to three ADL, "lying on a bed" (24 mis-detections), “getting in and out of a car seat” (19 mis-detections) and to a lesser extent, “sitting on an armchair” (10 mis-detections).

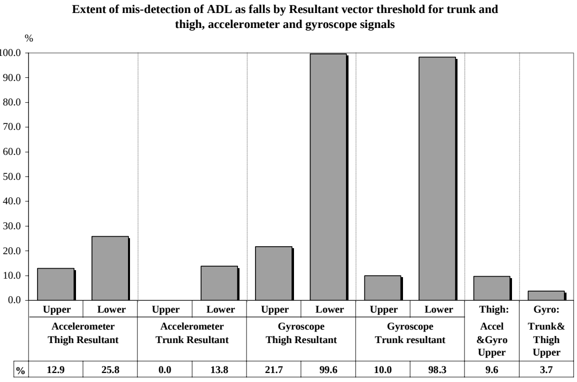
The single gyroscope axes were poor at distinguishing ADL from fall-events, with the Pitch axis, Upper threshold, producing the lowest mis-detection percentage on the trunk, mis-detecting 97.5% of ADL as fall-events, and the Roll axis, lower threshold, producing the lowest error percentage on the thigh, mis-detecting 98.3% of ADL as fall-events. Thus, the addition of single axis gyroscope sensors to the trunk and thigh did not complement the fall/ADL discrimination ability of single axes thresholds, Table 1.

Table 1: Extent of mis-detection of ADL as falls by single axis and combination threshold algorithms



The resultant vector signals from accelerometer and gyroscope sensors on the trunk and thigh were also analysed to determine the extent of mis-detected ADL as fall-events by these signals. The resultant vector trunk and thigh, accelerometer and gyroscope signals produced better fall mis-detection percentages than any of their single axes counterparts and within the resultant vector accuracies the upper threshold gave lower mis-detection error percentages then the corresponding lower thresholds, Table 2.

Table 2: Extent of mis-detection of ADL as falls by resultant vector threshold algorithms



With the Thigh accelerometer, resultant vector, Upper threshold, 12.9% of ADL were mis-detected as fall-events. The ADL “walking” and “getting in and out of a car seat” were responsible for 23 of the 31 (74.2%) mis-detected ADL. Including the “lying on a bed activity” and then these three activities were responsible for 26 of the 31 (83.3%) mis-detected ADL as falls.

In an important result using the trunk accelerometer, resultant vector, Upper threshold, 0% of ADL were mis-detected as fall-events, thus, no ADL were mistaken as fall-events from this sensors signal threshold, at the trunk.

With the thigh gyroscope, resultant vector, Upper threshold, 21.7% of ADL were mis-detected as fall-events. Again similarly to the thigh accelerometer, resultant vector, Upper threshold, the activities “walking” and “getting in and out of a car”, account for 30 of the 52 (57.7%) mis-detected ADL, with the “walking” ADL responsible for 19 of these 30. Adding the ADL, “lying on a bed” and “sitting on an armchair”, to this array of mis-detected ADL. Then these four contribute to 42 of the 52 (80.8%) of ADL mis-detected as fall-events.

Using the trunk resultant gyroscope Upper threshold, 10% of ADL were mis-detected as fall-events. The ADL that were mis-detected as falls were “lying on bed”, and, “getting in and out of a car seat”, which were responsible for 11 of 24 (45.8%) ADL mis-detected as fall-events. Including the activities “sitting on an armchair” and “sitting on a kitchen chair”, then these four ADL were responsible for 19 of the 24 (79.2%) mis-detected ADL.

The trunk and thigh resultant gyroscope Upper thresholds, were combined in an algorithm called “Gyro: Trunk & Thigh Upper”, using this algorithm 3.7% of ADL were mis-detected as fall-events. Four ADL were responsible for this error percentage, these were Sitting on an armchair”, “getting in and out of a car”, “sitting on a bed” and “sitting on a kitchen chair”, responsible for 3, 3, 2 and 1 mis-detections respectively.

Discussion

We have investigated single axis and resultant vector signals, from a tri-accelerometer and a bi-axial gyroscope placed at the trunk and thigh, to determine if differences in their peak values could be used to discriminate between an ADL and fall-events. In addition, a number of ANDed combination threshold algorithms were also investigated to determine their ADL/fall-event discrimination ability.

Even though the single axis gyroscope thresholds did not produce any reasonable accuracy on their own, with the pitch axis, Upper threshold, producing the lowest mis-detection percentage on the trunk, mis-detecting 97.5% of ADL as fall-events, the resultant gyroscope signal’s Upper thresholds, did produce low fall mis-detection percentages, 10% and 21.7% respectively.

Using the trunk tri-axial accelerometer, resultant vector, signal threshold (Upper threshold). 0% of ADL were mis-detected as fall-events. Thus, using this signal 100% of falls were correctly detected as fall-events and no ADL was mis-detected as fall-events, these percentages indicate that thresholding of this sensors signal is suitable, on it own, for incorporation into a primary fall-detection device

A possible limitation of the study was that young healthy males performed the simulated falls, however ethically it would be appropriate for elderly subjects to perform these falls. Also, in the simulated-fall study, the subjects fell onto large foam crash-mats. As the falls were performed onto a very soft surface, the faller's momentum was dissipated over a larger stopping distance as; the faller sank into the mat. Thus, the resulting peak impact force would not be as high as that experienced during an actual fall onto a solid surface. As a result, the signal from the accelerometer and possibly the gyroscope sensors also would have reduced peak values. Thus, the expected fall thresholds may in fact be higher, introducing an unintentional safety band in our fall thresholds.

Conclusion

In conclusion, we have investigated the use of single axis and resultant vector signals to detect fall-events, using thresholding of those signals, from tri-accelerometer and bi-axial gyroscope sensors placed at the trunk and thigh. A number of combination threshold algorithms, to accurately discriminate between ADL and fall-events were also investigated. We have shown that simulated falls can be distinguished from ADL using thresholding of the resultant accelerometer signal from a tri-axial accelerometer at the trunk with 0% mis-detection of ADL as fall-events (100% accuracy).

References

[1] TINETTI M. E., SPEECHLEY M., GINTER S.F. (1988): 'Risk factors for falls among elderly persons living in the community.' *New England Journal of Medicine.*, **319**, pp.1701-7

[2] SATTIN R. W., (1992): 'Falls among older persons: a public health perspective.' *Annual Review of Public Health*, **13**, pp. 489-508

[3] MAKI B., FERNIE G. (1996): 'Encyclopaedia of Gerontology: Age, aging and the aged Accidents: Falls. ' in BIRREN J. (ed), San Diego, Academic Press. 8, **1**

[4] HOYERT D. L., KOCHANEK K.D., MURPHY, S.L. (1999): 'Deaths: Final Data for 1997. National vital statistics reports', *Hyattsville, Maryland: National Center for Health Statistics*, **47**

[5] National Center for Health Statistics Vital Statistics System, (2000) Internet site address: http://www.cso.ie/

[6] TINETTI M. E., ET AL. (1994): 'Fear of falling and fall-related efficacy in relationship to functioning among community-living elders', *Journal of Gerontology*, **49**, pp. M140-47

[7] DOUGHTY K., (2000): 'Fall prevention and Managment strategies based on intelligent detection monitoring and assesment.' New Technologies in Medicine for the Elderly, Charing Cross Hospital, 30th. Nov. 2000

[8] DOUGHTY K., LEWIS R., MCINTOSH A. (2000): 'The design of a practical and reliable fall detector for community and institutional telecare.' *Journal of Telemedicine and Telecare*, **6**, S150-54

[9] HWANG J. Y., KANG J.M., JANG Y.W., KIM H.C., (2004): 'Development of Novel Algorithm and Real-time monitoring Ambulatory system using bluetooth module for fall detection in the elderly', Proceedings of the 26th Annual International Conference of the IEEE Eng. Med. Biol. Society, San Francisco, CA, USA. September 1-5, 2004

[10] DÍAZ A., PRADO M., ROA L.M., REINA-TOSINA J., SÁNCHEZ G., (2004): 'Preliminary evaluation of a full-time falling monitor for the elderly', Proceedings of the 26th Annual International Conference of the IEEE Eng. Med. Biol. Society, San Francisco, CA, USA. September 1-5, 2004., pp.2180-83.

[11] NOURY N., BARRALON G., VIRONE G., BOISSY P., HAMEL M., RUMEAU P., (2003): 'A smart sensor based on rules and its evaluation in daily routines', Proceedings of the 25th Annual International Conference of the IEEE *Eng. Med. Biol. Society*, Cancun, Mexico. September 17-21, 2003

[12] NI SCANAILL C., LYONS G.M., BOURKE A., KELLER M., SCHNIEDERS M., CULHANE K., AHERNE B. (2003): 'A GSM-based Remote Mobility Monitoring System for the Elderly', *ISSC*

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