# Evaluation of a fall detector based on accelerometers: a pilot study

U. Lindemann[1,2] A. Hock[3] M. Stuber[3] W. Keck[3] C. Becker[1]

[1]Robert-Bosch-Krankenhaus, Department of Geriatric Rehabilitation, Stuttgart, Germany
[2]Bethesda Geriatric Hospital Ulm, Academic Centre at the University of Ulm, Ulm, Germany
[3]University of Applied Science Ulm, Laboratory of Data Processing in Medicine, Ulm, Germany

**Abstract**—*As falls and fall-related injuries remain a major challenge in the public health domain, reliable and immediate detection of falls is important so that adequate medical support can be delivered. Available home alarm systems are placed on the hip, but have several shortcomings. A fall detector based on accelerometers and placed at head level was developed, as well as an algorithm able to distinguish between activities of daily living and simulated falls. Accelerometers were integrated into a hearing-aid housing, which was fixed behind the ear. The sensitivity of the fall detection was assessed by investigation into the acceleration patterns of the head of a young volunteer during intentional falls. The specificity was assessed by investigation into activities of daily living of the same volunteer. In addition, a healthy elderly woman (83 years) wore the sensor during the day. Three trigger thresholds were identified so that a fall could be recognised: the sum-vector of acceleration in the xy-plane higher than 2 g; the sum-vector of velocity of all spatial components right before the impact higher than 0.7 m $s^{-1}$; and the sum-vector of acceleration of all spatial components higher than 6 g. The algorithm was able to discriminate activities of daily living from intentional falls. Thus high sensitivity and specificity of the algorithm could be demonstrated that was better than in other fall detectors worn at the hip or wrist at the same stage of development.*

**Keywords**—*Accelerometers, Accidental falls, Fall detector*

Med. Biol. Eng. Comput., 2005, **43**, 548–551

## 1 Introduction

THE PREVENTION of falls and fall-related injuries in the elderly remains a major challenge in the public health domain as they result in physical, psychological and social limitations for the individuals and in enormous costs for the health care system (SKELTON and TODD, 2004). The reliable and immediate detection of falls is important so that adequate medical support can be delivered. Approximately 3% of all fallers lie for more than 20 min without external support (KING and TINETTI, 1995). This frequently leads to fear and social disengagement. Elderly, community-dwelling fallers have a high risk of being institutionalised even in the absence of severe injuries (TINETTI and WILLIAMS, 1997).

For the research community interested in fall prevention, the documentation of falls is a methodological pitfall. There is no unanimously accepted method for reporting falls. Data are scarce on the reliability and validity of different reporting systems. Oral reports are linked with many mistakes related to the cognitive status of the subjects, the shame of reporting, the fear of consequences and the work capacity of nurses. Falls associated with a loss of consciousness, such as syncope, stroke or epileptic seizure, are not always recognised. Currently, the documentation relies on oral report in subjects without cognitive impairment and proxy report in the elderly with dementia or other forms of cognitive impairment. KANTEN *et al.* (1993) demonstrated relevant differences in the reporting of falls in a nursing home setting. These differences introduce several sources of bias in observational as well as intervention studies (CUMMINGS *et al.*, 1988).

The available automatic home alarm systems are placed at the hip, but are not used widely and have several shortcomings. However, those few subjects wearing fall detector systems feel more confident and independent and consider that the detector improves their safety (BROWNSELL and HAWLEY, 2004). Furthermore, concerning non-automatic alarm systems, even in non-demented subjects, the elderly forget to use the alarm system in the event of a fall because they frequently panic.

Other systems in the development phase are worn at the wrist, but have a low sensitivity (65%) (DEGEN *et al.*, 2003). Several attempts have been made to improve home alarm systems with the use of sensor technology. Unfortunately, no scientific data are available on adherence and outcome, such as reduction in fear of falling. Not even the algorithm of fall detection is explained in one description of a fall detector (DOUGHTY *et al.*, 2000). Applicable technology includes, for example, gyroscopes and accelerometers.

Another consideration in the development of fall sensors is the site of signal detection. The detection at hip, trunk, wrist or head will result in different signal patterns leading to

*Correspondence should be addressed to Ulrich Lindemann; email: ulrich.lindemann@rbk.de*

*Paper received 16 March 2005 and in final form 27 June 2005*

*MBEC* online number: 20054044

different algorithms. This will influence sensitivity and specificity, which will highly influence adherence. A high rate of false alarms will be crucial for rejection of the system by the subjects and care givers. Lastly, it must be decided whether the impact of a fall, the incapacity to rise after a fall, or the fall itself is to be identified.

This study aimed to develop a device and an algorithm that could distinguish between activities of daily living and simulated falls. The sensor was fixed at the head to measure acceleration during activities of daily living and falling. It was assumed that the spatial and temporal patterns of movement of the head were more sensitive than those at trunk or hip sites to distinguish intentional and unintentional movements related to daily activities from falls.

## 2 Methods

Accelerometers were integrated into a hearing aid housing that was fixed behind the ear by a rubber sling. The rationale for placing the sensors on the head was the hypothesis that the individual tries to protect the head against high accelerations and that high accelerations measured at head level are associated with unpleasant movements such as falls.

Two $50g$ sensors* were used (for specifications see the Appendix) that convert the measured acceleration value into a proportional voltage value. Both two-axial sensors were placed orthogonally in the housing. Thus acceleration in all axes ($x$-axis = frontal, $y$-axis = sagittal, $z$-axis = vertical) was measured (Fig. 1).

The signals were digitised at a sampling rate of 200 Hz by a light, portable data logger† and stored for off-line elaboration on a memory card. The cable that connected the hearing aid and the logger was fixed by adhesive tape on the neck, near the ear, to prevent any tilt of the sensors caused by tension in the cable. The signals were filtered by a low-pass filter with a cutoff frequency of 80 Hz related to the sampling rate of 200 Hz. Acceleration as well as velocity (by first integration) could be derived from the signals.

The first step to derive an algorithm to detect falls was to identify typical fall patterns that were then performed by a young, healthy volunteer. The fall patterns were predefined by expert opinion based on patient reports in fall trials. The situations were designed to imitate typical fall situations reported by the elderly. The seven scenarios were a fall to the front, a fall to the side with a 90° turn, a fall to the back, a fall to the back with hip flexion, a fall backwards against a wall, imitation of a collapse, and a fall while picking up an object.

The sensitivity of the fall detection was assessed by investigation of the acceleration patterns of the head of a young volunteer during intentional falls. Each fall was conducted several times on a mat to avoid injuries to the volunteer. The specificity of the fall detection was assessed by investigation of activities of daily living in the same volunteer. The five manoeuvres chosen were sitting down on a chair, lying down, walking, running and stair climbing. Additionally, a healthy elderly woman (83 years) wore the sensor during the day. No fall occurred during this recording period.

At this stage of development, the system was applied only to a young volunteer (gymnast) and a healthy, active, elderly woman who had a low risk of falling. Persons of interest (i.e. elderly persons with a high risk of falling) were not included, because there could be a risk of their being injured by the logger during an unintentional fall.

*ADXL 250QC, Analog Devices, Norwood, USA
†Porti, Dr. Fenyvers und Gut, Hechingen, Germany

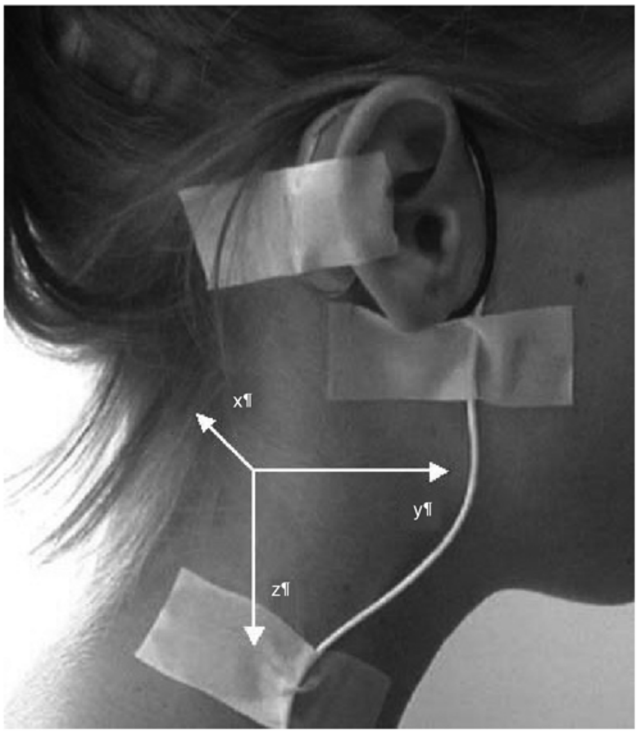


**Fig. 1** *Fall detector with co-ordinate system*

## 3 Results

No adverse events occurred while the fall detector and the logger were worn. The recording of the data of the intentional falls and the activities of daily living led to three trigger thresholds to recognise a fall. A fall was recognised when the sum-vector of acceleration in the $xy$-plane was higher than $2g$, and the calculated velocity (sum-vector of all axes) right before the initial contact was higher than $0.7\,\mathrm{m\,s^{-1}}$. In addition, a fall was recognised when the acceleration sum-vector of all spatial axes together was higher than $6g$.

With regard to the first threshold (sum-vector of acceleration in the $xy$-plane higher than $2g$), the $x$- and $y$-directions were analysed, because a change in the spatial direction of the head was observed, related to the eyes facing the ground during the fall ($y$-direction) or the side of the head touching the ground ($x$-direction). The $z$-direction was not included, because high acceleration values in this direction were measured during running and sitting down, which would have triggered a false alarm.

Concerning the second threshold (velocity right before the initial contact higher than $0.7\,\mathrm{m\,s^{-1}}$), it was assumed that 1500 ms after the initial contact with the ground, the faller's body was at rest, meaning the velocity of the head was 0. The value of 1500 ms was derived empirically, because, in all intentional falls, after 1500 ms there was no significant change of acceleration in all spatial axes. The velocity at the moment before the initial contact $V_0$ was calculated by backward integration of the acceleration $a$ of all spatial components from $t_1$ (1500 ms after initial contact) to $t_0$ (initial contact) (see (1) and Fig. 2). An integration constant was not included in (1), because velocity at $t_1$ was considered to be 0, thus returning a unique solution.

$$v_0 = \int_{t_1}^{t_0} a(t)\,dt \tag{1}$$

The rationale for looking at velocity and not at acceleration was derived from the fact that when the subject was laying his