A fall and loss of consciousness wearable detector

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Abstract— **Fall and Loss of Consciousness (FLoC) is the leading cause of serious health problems, above all in the elderly population, since the subjects involved are not able to ask for help and may lie in critical conditions for a long time thus deteriorating into severe conditions which can even lead to death. Therefore developing a device capable of automatically detecting a FLoC and activate an alarm call seems to be of utmost importance. The present study is intended to develop such a device using an accelerometer sensor. 460 simulated falls were performed by 20 subjects: 10 young subjects and 10 elderly subjects. The young subjects were asked to perform 200 FLoCs as well as 60 non common activities (NCAs), whereas the elderly subjects were asked to carry out only 200 activities of daily living (ADL). The signal used to detect the fall event was acquired by a single accelerometer placed on the subjects' belts. The test set was divided into two groups of the same size: Training Set (TS) and Verification Set (VS). The first set was meant to determine the related algorithm, whereas the second set was intended to check its reliability. The proposed algorithm was devised to detect the effects of the three phases of a FLoC (impact of the body on the ground, lying position and immobility) into the acceleration and jerk signals along the Cranio-Caudal Axis (CCA). The correct detection of all FLoC cases and the absence of false positives among ADL do corroborate the usefulness of the device proposed..**

I. Introduction

AMONG elderly people, falls with possible loss of consciousness are the leading cause of serious health problems. Age, gender, neurological disease [1] and limitation in walking performance [2] significantly influenced the occurrence of a fall. About 3.5 million Italians were victims of domestic injuries in 2000 [3]; falls represented the leading cause for 28% of the injuries in question and 60% of them had occurred at home.

Loss of consciousness can be a serious implication, above all for those elderly living on their own; it may either be the cause of the fall, as is the case for faints and epileptic fits, or result from head impact injuries, as is the case for traumatic brain injuries (TBIs). The victim of a fall and loss of consciousness is unable to ask for help and therefore likely to lie on the ground even for a prolonged period of time. This condition is particularly devastating when the "long-lie" phase- involuntarily remaining on the ground [4] - lasts more than one hour following a fall. In these cases, even when no serious injuries have occurred from the fall, half of the elderly involved die within less than six months of the occurrence [5].

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The commonest PERSs (personal emergency response systems) presently available to and widespread across the elderly living on their own, are provided with a voluntary push-button pendant. These systems are thus useless in FLoC cases. In recent years two different types of systems for the automatic detection of a FLoC have been reported in the literature. The first type consists of computer vision systems for home monitoring [6]. The second type consists of worn fall detectors that record the signal produced by tri-axial accelerometers [5, 7] and/or gyroscopes [8], mounted on several parts of the body that analyze different parameters [6, 9, 10, 11, 12, 13, 14]. In spite of this, only few hard data are available on the ability of accelerometry-based systems to detect falls in a community setting [15].

The present study has therefore been conducted to discriminate between FLoCs and the daily living activities (ADL) performed by elderly subjects, using a single accelerometer sensor based on the assumption that:

1. a FLoC is characterized by three phases:
   - impact of the body with the ground, caused by loss of balance;
   - misalignment between the Cranio-Caudal Axis (CCA) of the subject and the absolute reference Vertical Axis resulting from the body passing from an upright stance to a lying position;
   - "long-lie" condition as a result of loss of consciousness;
2. the effects of such phases on the accelerometer signal are, respectively:
   - peak values;
   - decrease in the mean value;
   - absence of changes;
3. a more efficient identification of the event can be obtained evaluating at the same time the effects described.

II. Materials and Methods

A. Data acquisition set-up

A prototype of the device [16]: a triaxial transducers (dimensions: 12 mm x 12 mm x 12 mm; weight 10 grams) and a wearable data-logger S.A.R.I. [17] developed in the authors' lab, was used for data acquisition. The accelerometric transducer was realized with two biaxial accelerometers ADXL210 (Analog Devices BV Ltd.), full scale range ± 10 g, (where g is gravity acceleration), sensitivity 250 mV/g, orthogonally mounted. A programmable controller unit (PIC16F877 Microchip Inc.) made it possible for the data-logger to i) acquire and analyze the signal through a low-cost computational algorithm, ii) monitor the push-buttons condition and iii) drive alarm systems (equipped with luminous, acoustic and vibrating alarms) and devices capable of activating emergency calls through GSM Network.

B. Participants

The study involved 20 subjects who were asked to perform 460 different movements: 10 young subjects (6 men and 4 women, age range: 33.6 ± 1.2 years; body mass: 70.4 ± 5.5 kg; and height: 1.75 ± 0.04 m) and 10 elderly subjects (age range: 75.8 ± 3.2 years,: body mass: 76.3 ± 7.8 kg; and height 1.68 ± 0.06 m).

The subjects, who were all in good state of health, gave their written informed consent before undergoing the test.

C. Test protocol

Both simulated FLoCs and non common activities (NCAs) took place in the authors' lab, whereas ADL were carried out in the performers' own homes, all under the supervision of a physical education professional. The young subjects performed 200 simulated FLoCs and 60 simulated non common activities NCAs, whereas the elderly subjects performed only 200 ADL. A single accelerometer was used, which was placed on the subjects' belts by means of elastic harnesses with Velcro snaps and correctly oriented [18].

The accelerometer signal was recorded with a 8 bits resolution, with a sample frequency of 100 Hz for 120 s to be transferred onto personal computer through S.A.R.I. RS-232 interface and processed using a dedicated programme developed in Matlab® (The Mathworks Inc., Natick, MA, USA).

Five different falls were performed:

- forward fall;
- slow forward fall;
- left lateral fall;
- right lateral fall;
- backward fall.

Each fall type was repeated four times. Three subjects performed the FLoC tests after hearing an acoustic signal in order to assess, with a certain time of reference, the start of the simulated fall and the duration of the impact phase.

All the other subjects were not given any acoustic signal for their falls to be more natural and closer to reality. Since 82% of falls usually occur when people are in an upright position [19], falls were simulated as occurring from standing height down onto a large crash mat (3 m x 2 m x 0.01 m).

Every subject was asked to walk at least five steps, to kneel down falling onto the crash mat to mimic real falls. The knee flexion phase took place at both normal and slowed-down speed to reduce ground impact. Before recording the signals, every subject performed test training.

The ADL performed by the elderly subjects included movements characterized by one or more elements typically present in a fall: peak accelerations, CCA rotation, immobility.

In more detail, each ADL consisted in:

- walking for 10 minutes;
- walking 5 steps and going down a step (height 0.10 m);
- walking 5 steps and sitting down a kitchen chair (height: 0.48 m);
- walking 5 steps and sitting down and lying down a bed (height: 0.54 m)
- walking 5 steps and bending forward to pick up an object from the floor.

Each ADL was performed 4 times by each subject.

The height of the step used for the testing (0.10 m) was lower than that of usual stairs and pavements (0.17-0.18 m) for a better simulation of the unevenness of the road pavements which is hardly perceivable and therefore particularly dangerous for the elderly.

The NCA tests were performed in order to reproduce all the typical elements of a FLoC (impact, lying position, immobility), and to assess the algorithm's ability to avoid false positives. Two movements were performed:

- walking 5 steps, going down a 0.10 m step, bending forward and remaining motionless with bent trunk for 60 s.
- walking 5 steps, turning aside, sitting down on a bed (height 0.54 m) and quickly lying down a bed and remaining motionless for 60 s.

Both movements were repeated 3 times at increasing speed.

D. Dataset partition

All trials were subdivided into two groups: Training Set (TS) and Verification Set (VS). TS was used to analyse

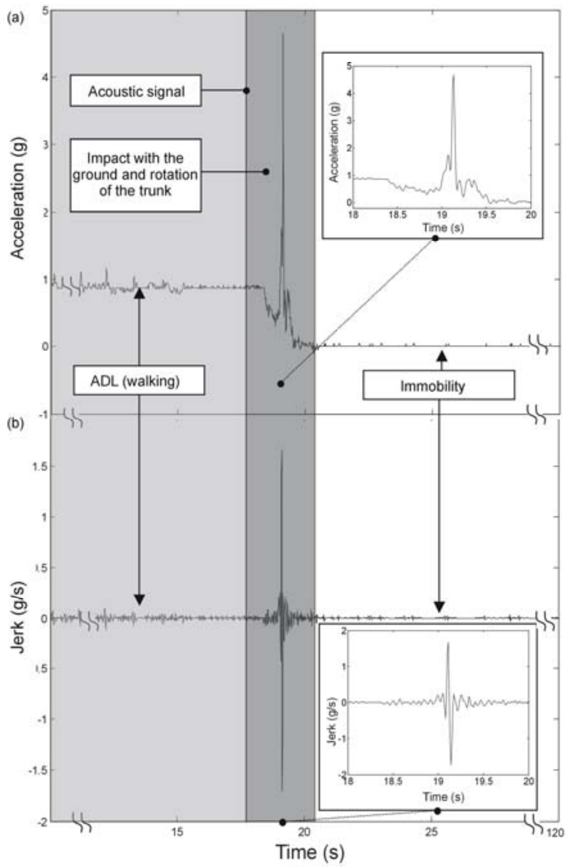

Fig. 1. typical features of FLoC detection signal. Acceleration (a) and jerk signals (b) along the CCA detect changes during the fall due to the impact of the subject with the ground (highlighted in the boxplots). In addition during the fall, the trunk rotates thus causing the gravitational component to change along the CCA. This change lasts throughout the immobility phase caused by the loss of consciousness when significant acceleration and jerk signal variations do not occur.

the signal trend and to set the signal threshold values; whereas VS was used to verify the accuracy of the detection algorithm. TS and VS had the same number of tests as well as the same percentages of FLoC, NCA and ADL.

E. Signal processing

A first evaluation of the accelerometer signals revealed the following:

- the CCA components had all the expected features of a FLoC (Fig.1);
- it was not possible to discriminate between FLoCs and ADL using the peak value neither of the acceleration modulus nor of its Cranio-Caudal Component (CCC), since in some cases such values may be higher in ADL than in FLoCs (Fig.2);
- the acceleration modulus was not capable of detecting the rotation of the trunk (Fig. 2);

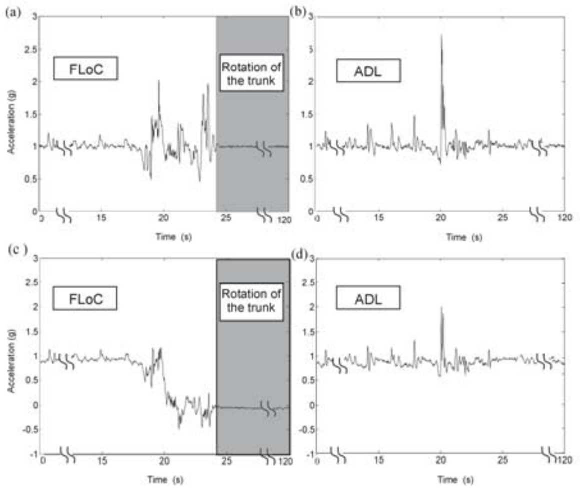

Fig. 2. the comparison between a) FLoC acceleration modulus and b) ADL - going down a step - acceleration modulus as well as the comparison between c) FLoC CCC and d) ADL - going down a step - CCC demonstrate that peak value does not allow to differentiate between FLoC and ADL; while comparing a) and c) it is clear that only CCC signal allows to detect the rotation of the trunk.

- immobility was clear in all signals.

FLoC detection was assumed to be determined by the identification of three interconnected conditions present in the signal: spikes (condition n.1) followed by a decrease in the mean component along the CCA (condition n. 2) and by concurrent absence of significant changes (condition n. 3). In some special conditions (e.g. epileptic fits) the latter third condition, might be absent. The detection of each condition was determined by comparing the data to the threshold value.

The duration of the impact phase was determined by analysing the triggered FloC signals. The start of the impact phase was set at 0.4 sec after trigger generation in order to take into account the subject's reaction time [21] .

The end of the impact phase was set as the instant in time after which there were no movements by the subject and therefore no significant variations in signal .

This condition was assessed by considering the last 10 sec of acquisition, with the subject lying still on the ground, and by calculating both mean and standard deviation (s.d.) values of the signal .

The end of the impact phase was defined as the instant in time after which the signal remained within the interval determined by the mean $\pm 3 \cdot$ s.d..

Since the effects due to both rotation of the CCA (condition n. 2) and immobility (condition n. 3) occurred and persisted after the impact phase in case of loss of consciousness, conditions 2 and 3 were checked simultaneously.

The verification procedure started when the impact phase was over, i.e. after a time interval equal to

$$T_2 = T_c + 3 \cdot \text{s.d.} \qquad (1)$$

where T_c and s.d. represent the mean value of the impact time and its standard deviation, respectively (the impact time had a positive response to the normality test). Moreover, prolonging these conditions for 60 s was believed to create a dangerous situation for the subjects since these events never last so long in ADL.

The impact phase was characterized by the presence of spikes. Hence in order to detect these spikes and evaluate the corresponding threshold value S0, the following parameters were analyzed:

- mean value of the acceleration CCC;
- mean value of rectified jerk CCC;
- peak value of rectified jerk CCC.

Each parameter was derived in successive time windows of amplitude $T_1 = T_c - 2 \cdot$ s.d.

It was suggested as follows:

- the impact took place at the same time of trunk rotation with both events producing a change in acceleration;
- after the impact phase the sensor, placed along the cranio-caudal axis, was no longer aligned to the vertical axis.

Therefore, after an interval time equal to T_2 from impact detection, the mean value of the acceleration CCC was obtained from a time window of 60 s. The lying position is assumed when this value is lower than the relevant threshold value S_1.

Subject's immobility caused a significant decrease in terms of acceleration changes; therefore the mean of the rectified jerk value was derived with the same procedure mentioned above. Immobility is assumed when this value is lower than the relevant threshold value S_2.

III. RESULTS

Both threshold values and arrangement of the signal processing procedure algorithm, defined by TS data, were evaluated according to the results obtained from VS data.

The duration of the impact phase exhibited a normal distribution (p = 0.489 – Anderson Darling) and its mean value T_c was equal to 4.4 s with a standard deviation of 1.7 s.

The three parameters analyzed were not normally distributed ($p < 0.005$ – Anderson Darling). Mean acceleration did not show statistically significant differences among the three activities simulated (Kruskal-Wallis $p = 0.613$); while the other two parameters enabled to clearly identify FloCs (Kruskal-Wallis $p < 0.0005$). The algorithm was devised in such a way so as to prioritize elimination of false negatives and reach 100% sensitivity ([1]). The mean value of rectified CCC jerk was selected in order to achieve the highest specificity ([2]), as shown in Table 1.

The algorithm was capable of correctly identifying the rotation of the trunk as well as the following immobility phase in all FLoC and ADL cases. Only 3% of NCAs CCA. Among the three parameters investigated, the mean value of rectified CCC jerk has turned out to be the most appropriate to detect the impact phase; also the immobility phase can be assessed by the same parameter, whereas the lying position is detected by the cranio-caudal acceleration trend.

The absence of false negatives demonstrates that the implemented algorithm and its hypotheses are clear enough. False positives have only been observed in NCAs, when rapid movements, differing from those made in “real situations”, have been performed.

Therefore it is possible to state that the device proposed, which relies on a single mono-axial sensor, is much more efficient and user-friendly (in terms of hardware, software

TABLE I.
MEAN AND THRESHOLD VALUES OF THE DETECTION PARAMETERS AND THE DETECTION RESULTS OF THE DIFFERENT CONDITIONS UNDER STUDY.

Event		Parameter	Mean (standard dev.)	Threshold values	FLoC Sensibility	ADL Specificity	NCA Specificity
Fall	**a)**	**Acceleration mean value**	0,27 (0,23) g	0,81 g	100%	7%	23%
	b)	**Rectified jerk peak**	$6{,}6\cdot10^{-3}$ $(4{,}1 * 10^{-3})$ g/s	$1{,}4\cdot10^{-3}$ g/s	100%	47%	33%
	c)	**Rectified jerk mean value**	$2{,}6\cdot10^{-4}$ $(0{,}9 * 10^{-4})$ g/s	$S_0 = 3{,}5\cdot10^{-4}$ g/s	100%	57%	75%
Lying position	**d)**	**Acceleration mean value**	0,02 (0,14) g	$S_1 = 0{,}35$ g	100%	100%	97%
Immobility	**e)**	**Rectified jerk mean value**	$5{,}1\cdot10^{-5}$ $(2{,}5*10^{-5})$ g/s	$S_2 = 11{,}0\cdot10^{-5}$ g/s	100%	100%	100%
All	**c) + d) + e)**		**TS**		**100%**	**100%**	**75%**
All	**c) + d) + e)**		**VS**		**100%**	**100%**	**63%**

The threshold fall detection values of the three conditions were devised through ROC curves [20], taking into account as first value the mean of the relevant parameter calculated in a time window of 6,1 sec (equal to Tc + ds) following the trigger instant and detecting the value corresponding to 100% sensitivity and highest specificity.

were false positives in terms of trunk rotation (Table 1).

The algorithm detected all FLoCs and ADL correctly using the three conditions proposed. The only false positives were registered in 25% of NCAs (Table 1).

As shown in Table 1 all FLoC and ADL tasks were detected correctly; the cases of false positives (37%) had to do only with NCAs. The values of all the parameters under consideration are given in Figure 3.

IV. CONCLUSION

The number of people likely to need emergency medical treatment without being able to ask for help -as is the case in FLoCs- is increasing as a consequence of the accelerated rate of ageing of the western population and of the growing number of people living on their own.

In this connection, a device capable of automatically activating an emergency call upon detection of a danger promises to be the ideal solution in terms of personal safety.

The results of the present work show that FLoC time domain characterization is possible by processing the changes in the accelerometer signal registered along the

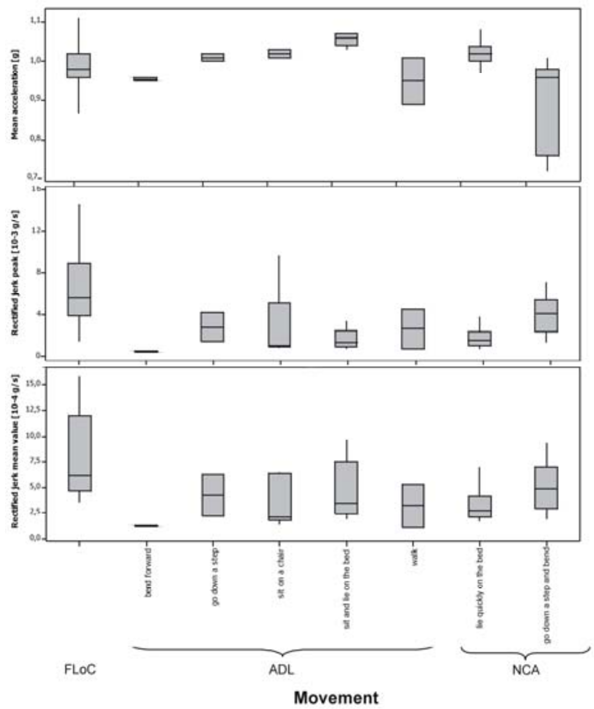

Fig. 3. The boxplot shows the values of the three parameters for all the trials of the same movement (FLoC, ADL, NCA). The signal of each trial was subdivided in time windows of 1 s and the highest of the mean values related to each time window was devised, in order to calculate mean acceleration and jerk.

[1] $T_{positive} / (T_{positive} + F_{negative})$ evaluated in FloC cases

[2] $T_{negative} / (T_{negative} + F_{positive})$ evaluated in all ADL and NCA cases

and general structure) than the devices previously reported in the literature.

The low computational cost of the algorithm in question allows for its implementation on a microcontroller with resulting hardware miniaturization (potential dimensions are expected to be around 7 cm x 4 cm x 2 cm with a weight of 250 g). The microcontroller allows to customize detection algorithm parameters and adapt them to the movements of the subject in question. Moreover, the low-power-consumption components used for signal acquisition and processing (about 10 mAh) allow the battery to last for at least 100 hours (with no alarm signal). The users should confine themselves to wearing and switch-on operations; before releasing the harness it should be switched off and charged, though it is designed to work for longer than the usual daily activity of elderly persons.

In addition, the device proposed, comprises an alarm system (active and passive), able to discriminate different situations and route emergency calls and further information through mobile phones or other devices in case of failure of was paid to realize active security systems. Passive security systems should be stout and ergonomic enough to be used while having a shower or a bath, i.e. in high-risk situations. Different kind of active alarm devices (acoustic, light-emitting and vibration) are planned. The detection of a FLoC generates an acoustic and vibratory pre-alarm (the latter is useful to subjects with hear impairment), while, if the detector is unfastened but not switched off, the fastening sensors (two tactile micro pushbuttons with low contact switching force, e.g. the type KSC401 - ITT Industries Inc. Santa Ana CA USA) are activated and the acoustic and light-emitting devices alert the subject.

In this work the FLoCs have been characterized by the concurrent presence of impacts, rotation of the trunk and immobility, but other kinds of falls may occur: slow FLoCs (i.e the subject slides from a lounge chair), FLoCs without rotation of the trunk (i.e the subject leans against a wall), FLoCs without immobility (i.e. epileptic attacks). In those cases, a threshold based detection system probably will not work properly. Therefore, more complex data processing systems are needed, such as soft computing algorithms.

Finally, the device proposed – for which a patent application has been submitted to the European Patent Office [22] – if appropriately adapted (e.g. for the transmission of the geographic coordinates of the location of the event in question), could also be used in other fields (sports, tourism, work activities) where individuals may find themselves in situations of isolation and risk.

References

[1] B. Resnick and P. Junlapeeya. "Falls in a community of older adults: findings and implications for practice," *Appl Nurs Res,*; vol. 17, no. 2, pp. 81-91, May 2004.

[2] D. C. Kerrigan, L. W. Lee, J. L. Collins, P. O. Riley, and L. A. Lipsitz. "Reduced hip extension during walking: healthy elderly and fallers versus young adults," *Arch Phys Med Rehabil*, vol. 82, pp. 26-30, Jan. 2001.

[3] ISTAT "Indagine multiscopo sulle famiglie "Aspetti della vita quotidiana" - Anno 2003". 2005.

[4] D. Wild, U. S Nayak, and B. Isaacs, "How dangerous are falls in old people at home?" *Br Med J (Clin Res Ed)*, vol. 282, pp. 266–8, Jun. 1981.

[5] A. K. Bourke, J. V. O'Brien, and G. M. Lyons, "Evaluation of a threshold-based tri-axial accelerometer fall detection algorithm," *Gait Posture* vol. 26, no. 2 pp. 194-199, Jul. 2007.

[6] H. Nait-Charif and S. J. McKenna, "Activity summarization and fall detection in a supportive home environment," in *Proc. 17th Intern. Conf. Pattern Recognition*, Cambridge, 2004; vol 4, pp. 323-326.

[7] 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 Comp*, vol. 43, no. 5, pp. 548-551, Sep. 2005.

[8] J. Y. Hwang, J. M. Kang, Y. W. Jang, and H. C. Kim, "Development of novel algorithm and real-time monitoring ambulatory system using Bluetooth module for fall detection in the elderly," *Proc. 26th Annu. Intern. Conf. EMBC*, San Francisco, CA, 2004; vol. 1, pp. 2204-2207.

[9] S. C. Jacobsen, T. J. Petelenz, and S. C. Peterson. U. S. Patent 6 160 478, Dec 12, 2000.

[10] J. M. Birnbach and S. D. Jorgensen, U. S. Patent Application 20 020 116 080, Feb 16, 2001.

[11] T. J. Petelenz, S. C. Peterson, and S. C. Jacobsen, U. S. Patent 6 433 690, Dec 11, 2002.

[12] D. Willis, "Ambulation Monitoring and Fall Detection System using Dynamic Belief Networks," Bc thesis. Monash Univ. Victoria, Australia, 2000.

[13] A. K. Bourke and G. M. Lyons, "A threshold-based fall-detection algorithm using a bi-axial gyroscope sensor," *Med Eng Phys*, vol. 30, no. 1, pp. 84-90, Jan, 2008.

[14] K. Doughty, R. Lewis, and A. McIntosh, "The design of a practical and reliable fall detector for community and institutional telecare," *J Telemed Telecare*, vol 6, no. 1, pp. S150–154, 2000.

[15] M. J. 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 Meas*, vol. 25: pp. R1–20, Apr. 2004.

[16] L. Quagliarella, N. Sasanelli, and G. Belgiovine, "An interactive fall and loss of consciousness detector system", *Gait Posture*; to be published.

[17] M. De Michele, N. Sasanelli, F. Attivissimo, V. D'annunzio, and L. Quagliarella, "Realization of a triaxial accelerometer for recording the fluidity of motion," *Gait Posture*, vol. 16: pp. S186, 2002.

[18] R. Moe-Nilssen, "A new method for evaluating motor control in gait under real-life environmental conditions. Part 1: The instrument", *Clin Biomech*, vol. 13, no. 4-5, pp. 320-327, Jun 1998.

[19] S. R. Lord, J. A. Ward, P. Williams, and K. J. Anstey, "An epidemiological study of falls in older community-dwelling women: the Randwick falls and fractures study," *Aust J Public Health*,vol. 17, no. 3, pp. 240–245, Sep. 1993.

[20] T. A. Lasko, J. G. Bhagwat, K. H. Zou, and L. Lucila Ohno-Machado, "The use of receiver operating characteristic curves in biomedical informatics", *J Biomed Informatics*, vol. 38, pp. 404–415, Oct. 2005.

[21] W. Consiglio, P. Driscoll, M. Witte, and W. P. Berg, "Effect of cellular telephone conversations and other potential interference on reaction time in a braking response," *Accident Analysis Prevention*,vol. 35, pp. 495–500, Jul. 2003.

[22] L. Quagliarella, N. Sasanelli, G. Belgiovine, and N. Cutrone. Eur Patent Application 06 425 414.7, Dec 26, 2007.