Accelerometry: providing an integrated, practical method for long-term, ambulatory monitoring of human movement

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TOPICAL REVIEW

Accelerometry: providing an integrated, practical method for long-term, ambulatory monitoring of human movement

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Abstract
Accelerometry offers a practical and low cost method of objectively monitoring human movements, and has particular applicability to the monitoring of free-living subjects. Accelerometers have been used to monitor a range of different movements, including gait, sit-to-stand transfers, postural sway and falls. They have also been used to measure physical activity levels and to identify and classify movements performed by subjects. This paper reviews the use of accelerometer-based systems in each of these areas. The scope and applicability of such systems in unsupervised monitoring of human movement are considered. The different systems and monitoring techniques can be integrated to provide a more comprehensive system that is suitable for measuring a range of different parameters in an unsupervised monitoring context with free-living subjects. An integrated approach is described in which a single, waist-mounted accelerometry system is used to monitor a range of different parameters of human movement in an unsupervised setting.

Keywords: accelerometer, ambulatory monitoring, falls, human movement

1. Introduction

Functional ability (or the restriction of it) is a determining factor in independent living. A person's level of functional ability has a significant impact on quality of life. Many tools have been developed to assist in the assessment of functional ability. These include photogrammetry, kinematic and kinetic analyses, video recording, electromyography, force plate analysis, simple-timed measures, questionnaire tools, validated functional tests and observation (see, for instance, Winters and Crago (2000), Adrian and Cooper (1989), Masdeu *et al* (1997)).

The difficulties with these approaches are that they are either time consuming and expensive, requiring access to specialized equipment and a dedicated laboratory set-up, or they are subjective and rely on clinician observation or patient recall. Moreover, measurements of movements made in the clinic may not accurately reflect functional ability in the patient's normal (home) environment (Kiani *et al* 1997).

There is a need for systems that are able to provide low-cost, objective measurement of functional ability of free-living subjects in the home environment.

A range of body-fixed sensors including electromechanical switches (attached to the heel to identify timing of heel strike in gait), goniometers (to measure joint angles between body segments), accelerometers (to measure motion of body segments), gyroscopes (to measure orientation of body segments), pedometers and actometers have been used to measure aspects of human movement in free-living subjects. Of these, accelerometers are becoming widely accepted as a useful tool for the assessment of human motion in clinical settings and free-living environments.

Accelerometers offer a number of desirable features in monitoring of human movement. Firstly, they respond to both frequency and intensity of movement, and so are superior to actometers or pedometers, which are attenuated by impact or tilt. Secondly, some types of accelerometers can be used to measure tilt as well as body movement, making them superior to motion sensors that have no ability to measure static characteristics. Thirdly, enhancements in microelectromechanical systems (MEMS) technology have made possible the manufacture of miniaturized, low cost accelerometers. These instruments also demonstrate a high degree of reliability in measurement, with little variation over time (Bouten *et al* 1997a, Meijer *et al* 1991, Moe-Nilssen 1998, Hansson *et al* 2001). This has enabled the development of small, lightweight, portable systems that can be worn by a free-living subject without impeding movement. Systems can be designed that are suitable for monitoring in the patient's normal environment over extended periods.

In this paper we investigate the potential for using accelerometry in this context. We begin by reviewing the work done in using accelerometers to measure various aspects of human movement. We then consider an integration of these systems and methods to produce a practical system for long-term monitoring of human movement in order to assess functional status in an unsupervised, free-living environment over extended periods.

2. Types of accelerometers

Accelerometers are instruments that measure the applied acceleration acting along a sensitive axis. There is a range of different transducers that are used to measure the acceleration. These include piezoelectric crystals, piezoresistive sensors, servo force balance transducers, electronic piezoelectric sensors and variable capacitance accelerometers. Although each type of accelerometer uses different mechanisms to measure the accelerations and there are many different designs and manufacturing techniques, conceptually, all use a variation of the spring mass system, shown in figure 1. In this system, when acceleration is applied, a small mass within the accelerometer responds by applying a force to a spring, causing it to stretch or compress. The displacement of the spring can be measured and used to calculate the applied acceleration. In terms of functionality, accelerometers can be divided into those that require an external power supply and those that do not, and into those that respond to static accelerations (such as the acceleration due to gravity) and those that do not.

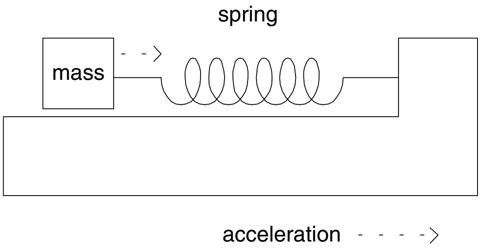

Figure 1. A piezoelectric accelerometer works like a simple mass spring system.

Most human movement applications use piezoresistive accelerometers or variable capacitance accelerometers (Mathie *et al* 2003b, Evans *et al* 1991, Murakami and Makikawa 1997, Veltink *et al* 1996, Sekine *et al* 2000, Fahrenberg *et al* 1997, Aminian *et al* 1995, 1999a), both of which respond to gravitational acceleration as well as to acceleration due to movement. Even if a static response is not required for the application, its presence allows the accelerometer to be calibrated by rotation within the gravitational field (Lötters *et al* 1998, Bouten *et al* 1997a). Piezoresistive sensors are typically made from a surface micromachined polysilicon structure on which polysilicon springs, whose electrical resistance changes with acceleration forces, are arranged in a Wheatstone bridge configuration. An applied acceleration produces a voltage proportional to the amplitude of the acceleration. Variable capacitance accelerometers typically use a differential capacitor with central plates attached to the moving mass and fixed external plates. An applied acceleration unbalances the capacitor, resulting in an output wave with amplitude proportional to the applied acceleration. Both of these types of accelerometers use MEMS technology, which results in them being miniature, low cost instruments. Both types require an external power supply that allows the accelerometer to respond to static accelerations.

More recently, accelerometers based on MEMS thermal sensor technology have become commercially available (Pyle and Emerald 2002). In these instruments, thermocouples are placed around a heating element to act like a Wheatstone bridge, where any difference in temperature between sensing elements results in a differential signal that is suitably amplified and conditioned. A change in acceleration results in a change in temperature gradient, and hence a change in output signal. These accelerometers also require an external power supply and can measure constant accelerations as well as changing accelerations. The advantage of this technology is that it allows for the cost effective measurement of accelerations with greater resolution than the previously described systems.

Neglecting any noise in the signal, if one of these accelerometers is held motionless then the measured acceleration is the projection of the gravitational acceleration vector onto the sensitive axis. If the instrument is moving then the measured acceleration is the vector sum of the gravitational and movement accelerations projected onto the sensitive axis (figure 2). Thus, the measured acceleration depends on both the movement and the orientation of the instrument relative to the gravitational field.

3. Monitoring human movements with an accelerometer

The position at which the accelerometer is placed on the body is important in the measurement of body movement. The accelerometer is normally attached to the part of the body whose movement is being studied. For example, accelerometers attached to the thigh or ankle are used to study leg movement during walking (Lafortune 1991, Bussmann *et al* 2000a, Aminian *et al* 1999a), accelerometers attached to the wrist have been used in the measure of Parkinsonian

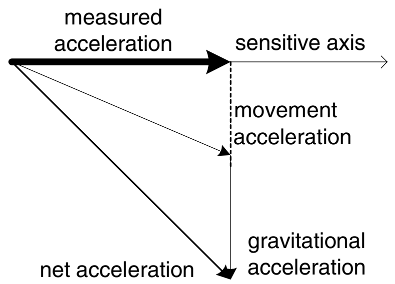

Figure 2. Piezoelectric accelerometers measure the sum of the acceleration due to movement and gravitational acceleration acting along the sensitive axis.

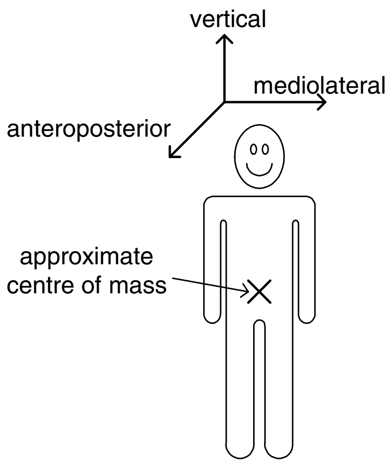

Figure 3. The centre of mass of a person is located within the pelvis when standing upright. The directions of the vertical, mediolateral and anteroposterior axes are shown.

bradykinesia (Veltink *et al* 1995), accelerometers have been attached to both arms and legs to study the Parkinsonian tremor (van Emmerik and Wagenaar 1996) and accelerometers attached to the chest have been used to study coughing (Fukakusa *et al* 1998). However, in many cases, investigators have wished to study 'whole body' movements. Some have achieved this by using multiple instruments placed across the body (Fahrenberg *et al* 1997, Foerster and Fahrenberg 2000, Uiterwaal *et al* 1998, Veltink *et al* 1996), while others have used a single instrument placed close to the centre of mass, which is located within the pelvis (Bouten *et al* 1997a, Smidt *et al* 1971, Sekine *et al* 2000).

The output of a body worn accelerometer depends on the position at which it is placed, its orientation relative to the subject, the posture of the subject and the activity being performed by the subject (figure 3). A uniaxial accelerometer records accelerations in a single direction. A triaxial accelerometer (TA) acts along three orthogonal axes and so can provide a picture of acceleration in three dimensions.

If the subject is at rest then the output of an accelerometer (with static response) is determined by its orientation relative to the gravitational vector. If the orientation of the accelerometer relative to the subject is known, then the accelerometer can be used to determine the orientation of the subject relative to the vertical (gravitational direction). If the subject is moving then the resulting signal is a combination of the subject's orientation and movements. This is illustrated in figure 4.

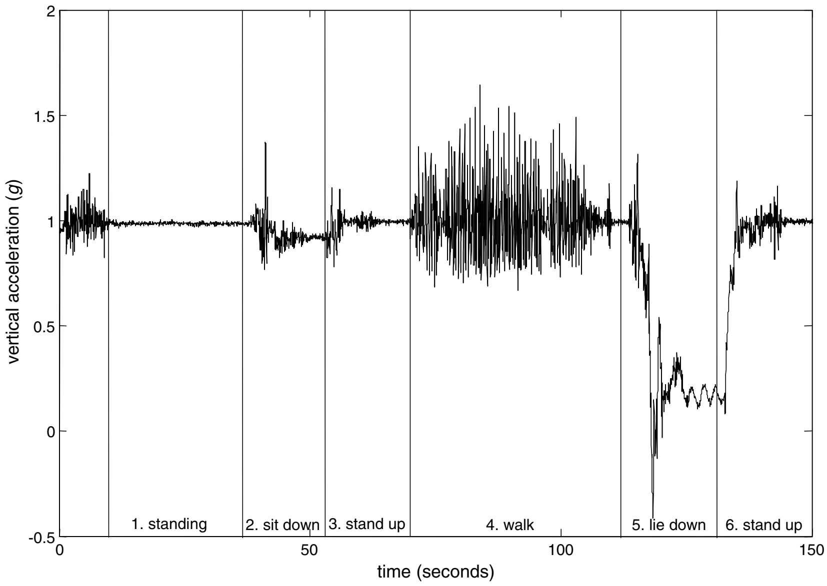

Figure 4. Acceleration signal produced by a waist-mounted accelerometer aligned in the vertical (gravitational) direction, during a selection of basic daily movements. The acceleration signal is composed of the gravitational acceleration due to the posture of the subject and the acceleration due to body movement. g is the acceleration due to gravity, approximately 9.81 m s^{-2}. The measured accelerations are dependent on the activity being performed. If the accelerometer was attached at a different point on the body, different acceleration signals would be recorded.

The movement component of the acceleration signal is itself composed of several different accelerations. The main components are due to movement of the body. Accelerations are generated due to both translational and rotational body movements. The relative contributions of the body-generated linear, centripetal and Coreolis accelerations to the overall signal are dependent upon the movement being undertaken.

Other accelerations, such as artefact due to soft tissue movement or external vibrations imposed on the body (due to travel in a motor vehicle) may also be present in the signal although these can be minimized through careful instrument placement and signal filtering.

Given that most human movements occur between 0.3 and 3.5 Hz (Sun and Hill 1993), most investigators have used a filter with a cut off frequency between 0.1 and 0.5 Hz to separate the two components of static orientation and body movement (Bouten *et al* 1997a, Fahrenberg *et al* 1997, Foerster and Fahrenberg 2000, Mathie *et al* 2003b, Veltink *et al* 1996).

4. Requirements for accelerometric monitoring of human movements

4.1. Acceleration amplitude and frequency range

The accelerations generated during human movement vary across the body and depend on the activity being performed. Accelerations increase in magnitude from the head to the ankle, and are generally greatest in the vertical direction (Bhattacharya *et al* 1980), although the accelerations in the other two directions cannot be neglected (Lafortune 1991).

Running produces the greatest vertical direction acceleration magnitudes, of 8.1–12.0g at the ankle (Lafortune 1991, Woodward and Cunningham 1993, Bhattacharya *et al* 1980), up to

5.0g at the low back (Bhattacharya *et al* 1980) and up to 4.0g at the head (Bhattacharya *et al* 1980). Walking down stairs also produces large accelerations, measuring up to 8.1g at the ankle (Woodward and Cunningham 1993). Trampoline jumping, walking up stairs, level walking and cycling produce lower acceleration magnitudes (up to 7.0g at the ankle during trampolining (Bhattacharya *et al* 1980), 7.4g during up stairs walking (Woodward and Cunningham 1993), and 2.9–3.7g during level walking (Lafortune 1991, Woodward and Cunningham 1993)). During walking, upper body accelerations in the vertical direction have been found to range from -0.3 to 0.8g, whereas horizontal accelerations range from -0.3 to 0.4g at the low back and from -0.2 to 0.2g at the head (Cappozzo 1982).

The major energy band for daily activities is 0.3–3.5 Hz (Sun and Hill 1993). Although foot acceleration at heel strike can reach frequencies of up to 60 Hz (Cappozzo 1982), 98% of the acceleration power during bare foot walking is contained below 10 Hz and 99% is contained below 15 Hz (Antonsson and Mann 1985, Aminian *et al* 1995). Slightly higher frequencies occur during running, but most acceleration is below 18 Hz at the ankle. The maximum frequencies obtained decrease from the ankle to the head, and are greater in the vertical direction than in the transverse plane (Cappozzo 1982).

In the light of such findings, Bouten *et al* (1997a) concluded that in order to assess daily physical activity, accelerometers must be able to measure accelerations up to $\pm 12g$ in general, and up to $\pm 6g$ if they are attached at waist level, and that they must also be able to measure frequencies between 0 and 20 Hz.

4.2. Designing a practical ambulatory monitor

There are design tradeoffs between the number of instruments that are used, the cost, the usability and the transferability of an ambulatory monitoring system. The design choices will be determined to a large extent by the purpose and duration of the monitoring. In short-term, supervised monitoring situations, large numbers of body-fixed sensors can be used to allow the collection of greater quantities of information, leading to very accurate assessments of movement. However, in long-term, unsupervised monitoring environments, subject compliance is essential if the system is to be used. In this situation, the wearable instrumentation needs to be easy-to-use, comfortable and as unobtrusive as possible. One approach is to embed multiple sensors into an item of clothing (Gallasch *et al* 1996, Park and Jayaraman 2003). The subject then has only to don the item of clothing, and all of the sensors are attached in the correct locations. However, increasing the number of sensors increases the complexity and cost of the system. Additionally, items of clothing must be designed in a range of sizes in order to ensure a proper fit on all subjects. A simpler approach is to use only one instrument that is attached at a single location on the body. This greatly simplifies the design and use of the system, but it also reduces the quantity of information that is obtained about the movements. A review of the literature demonstrates that, despite this limitation, useful information can in fact be obtained from a single device attached near the centre of mass of the subject (see, for example, Mayagoitia *et al* (2002), Sekine *et al* (2002), Bouten *et al* (1997a), Evans *et al* (1991)).

5. Capabilities of accelerometry in monitoring human movement

5.1. What is important in unsupervised monitoring of human movement?

There are three main purposes for which an ambulatory monitoring system can be used. Firstly, it can be used for objective assessment of particular movements. For example, an ambulatory monitoring system may be used during a clinical assessment of gait. Secondly, it can be used to monitor for adverse events, such as falls, during unsupervised free living. Thirdly, it can be used to monitor important movements during unsupervised free living, and to allow longitudinal tracking of important parameters. Important movements include those that are predictive of falls, and those that are indicative of functional status. Measurements of, for instance, postural sway, walking speed and step rate variability provide information on risk of falling (Lord and Clark 1996, Hausdorff *et al* 1997, Fernie *et al* 1982, Luukinen *et al* 1995, Kamen *et al* 1998, Guimaraes and Isaacs 1980). Movements such as the sit-to-stand transfer and walking are important for functional independence (Munro *et al* 1998, Kerr *et al* 1997). These movements, and the amount of time spent in activity each day, provide important information on functional status particularly when they are assessed in the home environment. Finally, if the system is to be used in an unsupervised setting to monitor movements of interest then it also needs to have an ability to identify these movements in the signals produced by the monitoring system.

Studies have been performed in which accelerometers have been used to assess many of these important movements and parameters, including metabolic energy expenditure, physical activity, postural sway, gait, falls detection, and postural orientation and activity classification. Each of these areas is reviewed below.

5.2. Metabolic energy expenditure

The standard reference for the measurement of physical activity is the metabolic energy expended due to that physical activity (Servais and Webster 1984, Bouten *et al* 1997a). Metabolic energy expenditure (EE) can be measured relatively easily in a laboratory using direct calorimetry, but is difficult to measure in free-living humans. Accelerometry provides an indirect method for assessment of physical activity suitable for free-living environments that performs favourably when compared with other indirect methods such as self-reporting (Steele *et al* 2000, Ng and Kent-Braun 1997) and heart rate measurements (Fehling *et al* 1999, Bussmann *et al* 2000b).

Accelerometers have been used to study physical activity and energy expenditure in populations including healthy young subjects (Bouten *et al* 1996, Miller *et al* 1994), elderly subjects (Meijer *et al* 2001, Kochersberger *et al* 1996), patients with multiple sclerosis (Ng and Kent-Braun 1997), patients with chronic obstructive pulmonary disease (COPD) (Steele *et al* 2000) and obese children (Coleman *et al* 1997, Epstein *et al* 1996, Trost *et al* 2001). Instruments that use accelerometry to measure EE or physical activity levels are available commercially (for example, Caltrac and Tritrac-R3D (Madison, WI) and the computer science and applications (CSA) 7164 activity monitor, (Shalimar, FL)).

Accelerometry systems typically use a model in which the area under the curve traced by the time course of body movement acceleration is linearly related to EE; i.e. the acceleration signal magnitude area is proportional to energy expenditure (figure 5). This relationship has been demonstrated for uniaxial accelerometers (Montoye *et al* 1983) and for triaxial accelerometers (Bouten *et al* 1997a, 1997b, Steele *et al* 2000). In a back-to-back comparison of 11 estimators derived from a waist-mounted triaxial accelerometer, Bouten *et al* (1994) found that the integral of the magnitude of the acceleration in the antero-posterior direction was the best estimator of energy expenditure during walking and the best estimator across a range of daily activities was the sum of the integrals of the magnitudes of each of the three accelerations. The actual energy expenditure was then calculated using a linear regression. Chen and Sun (1997) investigated the use of a nonlinear regression model and reported that it gave significantly more accurate results than the linear model. They also found that body

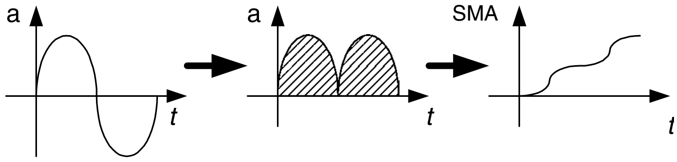

Figure 5. Metabolic energy expenditure (EE) is estimated by computing the signal magnitude area (SMA). The acceleration signal is rectified and then integrated. EE is then estimated by means of a linear regression.

mass was a significant factor in the relationship between the acceleration signal and the energy expenditure.

The use of three-dimensional acceleration significantly improves the accuracy of the EE estimate over that made using acceleration in only one dimension (Ayen and Montoye 1988). Adding additional instruments at different locations on the body, for example, at the wrist, provides only a very slight improvement in accuracy, and does not justify the extra cost, complexity or inconvenience caused by the addition of a second instrument (Bouten *et al* 1997b, Swartz *et al* 2000).

The placement and orientation of the TA device on the body appear to have a negligible effect on the correlation between the accelerometer estimated EE and the measured EE. Bouten *et al* (1997b) measured accelerometer output in the antero-posterior and vertical directions at the low back, and from this simulated accelerations at the shin, upper leg, trunk, lower arm and upper arm. The acceleration-derived EE measures calculated from these simulated values were compared to the metabolic energy expenditure. They concluded that they were able to use the accelerometer output at all examined locations to predict the EE with a high degree of accuracy.

The accuracy of the accelerometer EE estimate depends on the activity being performed (Fehling *et al* 1999, Hendelman *et al* 2000). The estimate from a waist-mounted accelerometer is accurate during walking on a level surface (Bouten *et al* 1994, Terrier *et al* 2001), but activity that is concentrated in the upper body, such as weight lifting, or washing the dishes is significantly underestimated by a waist-mounted accelerometer (Hendelman *et al* 2000, Bouten *et al* 1994). Accelerometers cannot measure the energy cost of walking up or down a slope, compared to walking on level ground (Terrier *et al* 2001), although use of additional processing techniques, such as artificial neural networks, have shown success in estimating the incline along which a subject is walking (Aminian *et al* 1995), and this information may be able to be used to provide improved estimation of EE.

However, the main advantage of using an accelerometer to measure EE is that it can be used in general free-living conditions. The accuracy of the EE estimate has been shown to vary greatly under free-living conditions, depending on the activities that were being performed. Bouten *et al* (1997a) reported a pooled correlation coefficient of 0.89 for 13 subjects over a 36-h period spent in a house-like chamber where they performed standardized daily activities. On the other hand, Hendelman *et al* (2000) reported correlations of only 0.59–0.62 for subjects completing a range of indoor and outdoor activities including playing golf, vacuuming and lawn mowing. They attributed this to the inability of accelerometers to detect increased energy cost from upper body movement, load carriage, or changes in surface or terrain.

5.3. Physical activity

Closely associated with the measurement of metabolic energy expenditure is the use of accelerometry to assess physical activity and to compare physical activity levels between groups of subjects. Accelerometers allow the collection of objective measurements that are proportional to activity intensity in free living.

Both uniaxial accelerometers (such as the CSA) and triaxial accelerometers (such as the TriTrac-R3D) have been attached at the waist and used to assess physical activity. In the CSA, vertical accelerations are sampled at 10 Hz, filtered, digitized and summed over a user-selected epoch to produce an activity count per epoch (Tryon and Williams 1996). The Tritrac-R3D accelerometer uses an almost identical approach, except that acceleration counts from three dimensions are summed to provide an activity level count (Steele *et al* 2000).

The uniaxial and triaxial accelerometers have been shown to provide reliable, valid and stable measurements of physical activity levels when compared to other indicators of functional capacity. They have been used to study physical activity in many different cohorts, including groups of children (Cooper *et al* 2003, Trost *et al* 2001), youth (Pate *et al* 2002, Trost *et al* 2002), obese populations (Cooper *et al* 2000), COPD patients (Steele *et al* 2000), patients with multiple sclerosis (Ng and Kent-Braun 1997) and the elderly subjects (Kochersberger *et al* 1996).

Consistent with the results reported in the preceding section on metabolic energy expenditure, Freedson *et al* found a correlation of 0.86 between predicted and observed MET level using a regression between METs, age and CSA activity counts/min (reported by Trost *et al* (2001)). Ekelund *et al* (2001) found that activity counts from the CSA activity monitor were a useful measure of the total amount of physical activity in nine year old children and that activity counts contributed significantly to the explained variation in total energy expenditure ($r = 0.39$; $P < 0.05$) and in activity energy expenditure ($r = 0.54$; $P < 0.01$).

Most studies using accelerometers for assessment of physical activity have used a measure of counts per minute (typically averaged hourly) over a study period between three days and two weeks in order to assess physical activity (Cooper *et al* 2000, 2003, Ekelund *et al* 2001, Pate *et al* 2002, Steele *et al* 2000, Trost *et al* 2002).

5.4. Balance and postural sway

Postural stability is the ability of an individual to maintain the position of the body within specific boundaries of space without needing to move the base of support. This requires the complex integration of sensory information regarding the position of the body relative to the surroundings, and the ability to generate forces to control the body movement. Measurement of postural sway during quiet and perturbed standing is often used to assess balance and falls risk (Sherrington and Lord 1998, Lord and Clark 1996, Campbell *et al* 1989).

Both amplitude and frequency are important in assessment of postural sway, with large sway amplitudes and higher frequencies being indicative of postural instability (Fernie *et al* 1982, Kamen *et al* 1998). It has been postulated that the harmonic content of the postural sway signal contains information regarding the degeneration of the balance control system due to ageing and balance-related pathologies (Winter 1995). The spectral pattern of sway obtained using a force platform has been found to be useful in distinguishing between various pathological conditions, and it has been suggested that quantitative sway assessment may be most important in the identification of subtle and idiopathic falling disorders (Kamen *et al* 1998).

Traditional tests of postural sway (including the Romberg test, the Wright ataxiometer and the force platform analysis) need to be set up and conducted by an observer, not by the patient him or herself and so are not suitable for assessment of balance during routine daily activities, nor for continuous monitoring. Accelerometry has been found to be a reliable method for measurement of balance during standing and walking, with high absolute test–retest reliability (Moe-Nilssen 1998). The signals obtained from a sacrum-mounted accelerometer can be used to distinguish between different balance conditions including standing on firm and compliant surfaces, standing with feet together and feet apart, and standing with eyes open and eyes closed (Cho and Kamen 1998, Mayagoitia *et al* 1999, 2002, Kamen *et al* 1998). Accelerometric variables measured during quiet standing can also be used to distinguish between healthy elderly subjects and idiopathic elderly fallers (Cho and Kamen 1998).

5.5. Gait

In addition to being an important skill for independent living, parameters of gait can provide indication of deteriorating functional ability and increasing falls risk. Walking speed is related to functional status (Friedman *et al* 1988, Potter *et al* 1995), and is a predictor of falls (Guimaraes and Isaacs 1980, Luukinen *et al* 1995). Fallers also walk with greater variability in stride time, stance time and swing time than non-fallers (Hausdorff *et al* 1997, Lord *et al* 1996, Cho and Kamen 1998).

It has been shown that simple parameters such as step and cycle time and stride symmetry can be determined during normal gait from waist, thigh or heel accelerations (Currie *et al* 1992, Evans *et al* 1991, Foerster and Fahrenberg 2000, Terrier *et al* 2000, Bussmann *et al* 2000a, Auvinet *et al* 1999). Accelerometers attached to the legs have been used to enable automated extraction of temporal gait patterns including left and right heelstrikes and toe-offs (Aminian *et al* 1999a). Aminian *et al* (1995) used two neural networks to estimate incline and walking speed during unconstrained walking using a triaxial accelerometer attached to the back and a uniaxial accelerometer attached to the top of the right heel. The standard deviation of the estimated incline was less than 2.6%, and the maximum of the coefficient of variation between speed estimation was 6%. However, after applying a similar approach, Herren *et al* (1999) reported that their system allowed accurate prediction of speed but not of incline during running.

Outdoor walking speed has been accurately measured using a combination of accelerometry and altimetry (Perrin *et al* 2000).

Sekine *et al* (1998, 2000, 2002) and Tamura *et al* (1997) have demonstrated that walking on level ground and walking on a stairway can be distinguished in the signals of a waist-mounted triaxial accelerometer.

The vertical acceleration component of the trunk- or back-mounted TA is the most important in the assessment of gait (Bouten *et al* 1997a, Fahrenberg *et al* 1997, Foerster and Fahrenberg 2000). This is the component that is most sensitive to the presence of gait disorders (Liberson 1965), and from which elements of the gait cycle can most easily be identified (Evans *et al* 1991).

Smidt *et al* (1971) defined a measure of smoothness of walking, called the harmonic ratio as the sum of the coefficients for the even-numbered harmonics of the Fourier series, divided by the sum of the coefficients for the odd-numbered harmonics; the greater the harmonic ratio, the smoother the walking. The investigators found that this harmonic ratio provided an effective method for discriminating between normal gait patterns and gait patterns of subjects with gait defects. Farris *et al* (1993) similarly found that symmetry in gait could be seen in the prominence of even harmonics in the accelerographic signal.

Preliminary research suggests that the power spectrum of the accelerometer signal can also be used to assess the stability of gait. In an initial study, a performance parameter based on the balancing forces as reflected in the power spectrum was successfully used to order different gait patterns in terms of relative stability (Waarsing *et al* 1997).

5.6. *Sit-to-stand transfers*

The ability to rise from a chair is of fundamental importance for functional independence. Rising from a chair is regarded as the most mechanically demanding functional task undertaken during daily activities (Kerr *et al* 1997) and is a prerequisite for gait (Kralj *et al* 1990). An inability to rise from a chair can prevent an otherwise functionally independent subject from independent living (Munro *et al* 1998). The ability to sit down in a controlled manner is of equal importance.

Little work has been reported using accelerometers for assessment of the sit–stand–sit movement. Sit-to-stand and stand-to-sit transitions can be automatically identified as periods of activity (Mathie *et al* 2003b), and they can be classified by identifying the preceding and succeeding postures as sitting and standing (Aminian *et al* 1999b, Fahrenberg *et al* 1997). A preliminary study found a moderate correlation ($r = 0.537$) between the accelerometry characteristics of the sit-to-stand transfer measured at the waist and falls risk in 37 elderly subjects (Troy *et al* 1999). Other useful clinical information may be able to be obtained from the accelerometry signals of the sit–stand–sit movement, but this remains to be investigated.

5.7. *Falls*

One of the biggest risks to the health and well being of the elderly is the risk of morbidity from injury, leading to functional dependence. Falls are a very serious risk for the elderly, particularly for those living in the community. In those aged over 65 years, two thirds of accidents are falls (Hawranik 1991) and, for example, in the general Australian community, accidents are the fifth leading cause of death, and one quarter of them are falls (Australian Bureau of Statistics 1997).

Accelerometry has been proposed as being suitable for falls detection in free-living subjects but there has been relatively little work done in this field to validate the method. The basic approach was first published by Williams and Doughty (Williams *et al* 1998, Doughty *et al* 2000). In this approach, a change in orientation from upright to lying that occurs immediately after an abrupt, large negative acceleration (due to impact) is indicative of a fall. Both of these conditions can be detected using an accelerometer that has a dc response, and have been incorporated into fall detection algorithms using an accelerometer (Jacobsen *et al* 2000, Birnbach and Jorgensen 2002, Petelenz *et al* 2002, Lehrman *et al* 2002, Mathie *et al* 2001, Salleh *et al* 2000).

However, little real data are available on the ability of an accelerometry-based system to detect falls in a community setting. This remains an area requiring further work.

5.8. *Movement classification*

Accelerometry systems have been used to identify and classify sets of postures and activities. Most of these systems have used multiple sensors; some systems have used only accelerometers, while other systems have used accelerometers together with another type of sensor. The most common placement locations are the chest or waist and the thigh (Aminian *et al* 1999b, Uiterwaal *et al* 1998, Fahrenberg *et al* 1997, Foerster and Fahrenberg 2000, Veltink *et al* 1996, Bussmann *et al* 1998).

Algorithms for the detection of posture and motion patterns remain a crucial aspect of accelerometry, and the ability to achieve an adequate data reduction while still being able to differentiate between a variety of dynamic activities is still under investigation (Foerster and Fahrenberg 2000).

Classification systems have used fixed-threshold classification methods using empirically derived thresholds (Mathie *et al* 2003b, Sekine *et al* 2000, Aminian *et al* 1999b), pattern-matching techniques in which the signal is compared to fixed template reference patterns (Murakami and Makikawa 1997, Makikawa and Murakami 1996), statistical-based algorithms (Veltink *et al* 1996, Fahrenberg *et al* 1997), fuzzy logic (Winters *et al* 2003), and artificial neural networks (Aminian *et al* 1995, Kiani *et al* 1997). Often, systems have used combinations of techniques.

Systems have been developed to identify the postural orientation of a subject (Pianca *et al* 2002, Sheldon 1997). Other systems have used accelerometers placed on the chest or waist and the thigh to discriminate between postures and activities sitting, lying, standing, walking, stair climbing and cycling with a high degree of accuracy (Aminian *et al* 1999b, Uiterwaal *et al* 1998, Veltink *et al* 1996, Bussmann *et al* 1998, Kiani *et al* 1997) by first discriminating between activity and rest, and then between different resting postures, and different activities. Accelerometry systems using multiple instruments placed across the body have been also used to achieve classification of multiple activities and postures (Fahrenberg *et al* 1997, Foerster and Fahrenberg 2000, Bussmann *et al* 2001). Accelerometry has also been used in conjunction with heart rate (Makikawa and Iizumi 1995), GPS (Makikawa and Murakami 1996) or gyroscopes (Luinge *et al* 1999) to classify postures and activities.

A range of classifications can also be achieved using a single instrument. Mathie *et al* (2003b) demonstrated that periods of activity and periods of rest could be reliably separated using only the signals from a waist-mounted triaxial accelerometer. Using the same instrument, they also found that there were statistically significant differences between the acceleration signals obtained during periods of standing, sitting and lying (Mathie *et al* 2003c). Murakami and Makikawa (1997) used a biaxial, waist-mounted accelerometer to classify free movements as standing, walking, sitting, lying and travelling by vehicle.

The majority of movement classification systems have been custom designed for a specific domain of postures and activities. Although many of these systems have produced excellent results in classification of specific movements, there is still scope for the development of systems that are able to automatically identify and classify arbitrary movements performed in free-living conditions.

6. Monitoring in an unsupervised home environment

The studies that have been undertaken have demonstrated the utility of using accelerometers for monitoring human movements and quantitatively measuring important parameters of movement. The next phase in the research and development of accelerometric monitoring systems needs to focus on the application of these devices for unsupervised monitoring of free-living subjects. The technical requirements for monitoring in the home environment must be addressed, including instrument usability, power supply, reliable wireless communications and secure transfer of information. Many of these issues are being resolved with the development of home wireless network technologies and very low power instruments that are designed to be used in wearable monitoring systems (Asada *et al* 2003, Celler *et al* 2000, Jovanov *et al* 2003, Korhonen *et al* 2003). Additionally, expectations of clinical outcomes must be addressed in terms of the monitoring instrument's known capabilities, and algorithms that function robustly in free-living conditions must be designed.

The design requirements depend on the intended purpose of the monitoring system. These are discussed in the following section.

6.1. Three modes of operation

When an accelerometry system is to be used for monitoring a free-living subject, there are three basic modes in which it can be applied. These are clinical assessment, event monitoring and longitudinal monitoring. Each mode of monitoring fulfils a different function, and the system is used differently in each case.

6.1.1. Clinical assessment. The accelerometry system can be used to provide a clinical assessment of the subject in the home environment. This typically requires a one-off performance of a series of known tasks. The assessment will normally, although not necessarily, be performed in the presence of a health professional and the objective, quantitative information from the accelerometry system can be used to supplement qualitative assessment.

This one-off, supervised monitoring mode differs from the other two modes, which both involve long-term, unsupervised monitoring.

6.1.2. Event monitoring. The accelerometry system may be used to monitor for abnormal events, such as falls or long periods without movement. In this mode the system functions as a smart personal alarm system. Continuous monitoring of the subject and real-time data processing are needed in order to detect abnormal events and raise the alarm in a timely manner. In this mode, the system requires a reliable connection to an emergency call centre or carer, via the phone network, or any other means, so that contact can be made in the event of an emergency, regardless of the location of the subject in the home.

6.1.3. Longitudinal monitoring. The accelerometry system can also be used to monitor parameters of movement that are sensitive to changes in health or functional status, such as those that have been described in previous sections.

The parameters to be monitored can be subdivided into two basic types: those that are movement specific and those that pertain to the movement generally. Movement-specific parameters include step rate, postural sway and rise time. Before these parameters can be measured, a particular movement needs to be performed and identified. These parameters are not continuously monitored because subjects do not continuously perform the requisite movements. General parameters of movement include hourly and daily physical activity and metabolic energy expenditure, and the amount of time spent resting during the day. Continuous monitoring is required to compute these parameters.

Before movement-specific parameters can be measured, the movements must be identified. This can be achieved by applying classification algorithms to the signals obtained during the free movement in order to identify movements of interest. Alternatively, the subject can be directed to carry out a particular movement. In this latter case, the movement is known *a priori* and no classification is required. This approach greatly reduces the processing that is required, as continuous processing is not necessary, but only processing of the signals derived from the directed movements. This is appropriate for general monitoring of functional status. A routine of directed movements can be composed and undertaken by the subject at regular intervals, for example, daily or weekly. A major advantage of this method is that the same movements are repeated in the same manner each time that the routine is carried out, and this allows direct comparison of the movements over time. In free-living conditions, subjects carry out the same movement in different manners on different occasions, depending on circumstance and this can dramatically change the resulting acceleration patterns. Use of a directed routine helps to reduce the variability in the movements. On the other hand, this approach is not adequate for monitoring in instances where rapid changes in functional status may occur, as in the case

Table 1. Summary of the requirements for each of the three modes of monitoring.

Function	Clinical assessment	Event monitoring	Longitudinal monitoring: Specific parameters of movement	Longitudinal monitoring: General parameters of movement
Continuous monitoring required?	No	Yes	No	Yes
Repeated measures required?	No	No	Yes	No
Activity classification required?	All relevant activities need to be classified	Falls and other abnormal events need to be detected	All relevant activities need to be classified	Depends on parameters to be measured. Not required for EE
Longitudinal tracking required?	No	No	Yes	Yes
Alerting/alarming functionality required?	No	Yes	Yes	Yes
Timing critical?	No	Yes	No	No
Purpose	Clinical assessment	Respond to emergency	Predictive/preventative	Predictive/preventative

of a medication that may lead to syncope and instability for a short period after ingestion. For this type of situation, continuous monitoring is more appropriate, despite the increased complexities in data processing.

Longitudinal changes in parameters should be flagged and an appropriate carer or health care worker alerted. The three modes of monitoring are summarized and compared in table 1.

6.2. *An integrated approach*

It is possible to integrate these different modes of monitoring to produce a single accelerometry-based system that can be used to supplement clinical assessment in the home; to longitudinally track a range of important parameters of movement in order to identify changes in health or functional status; and to monitor for emergency events, such as falls.

We have considered the problem of providing unsupervised, long-term monitoring to frail elderly, housebound patients. For this group of people, the focus of the system should be on providing health support to the subject to detect problems and to attempt to prevent morbidity through early detection, leading to timely intervention (Celler *et al* 1999).

Consequently, we designed a system that was intended to operate in the last two monitoring modes, that of event monitoring and longitudinal tracking (Mathie *et al* 2000, 2001). Similar schemes have also been investigated by Kiani *et al* (1997). The wearable instrumentation consisted of a single, waist-mounted triaxial accelerometer. A single instrument was used to facilitate use and to minimize the cost of the system. In the prototype system, the wearable triaxial accelerometer unit sampled accelerations at 45 Hz and transmitted the data via a 433.92 MHz wireless link to a personal computer, where each sample was time stamped, processed and stored. The instrument is able to transmit over a range of up to 50 m line-of-sight, and the radio frequency transmission means that the signals can ‘bend’ around walls in the home. One 1.5 V AA alkaline battery provides the unit with sufficient power for 80 h of

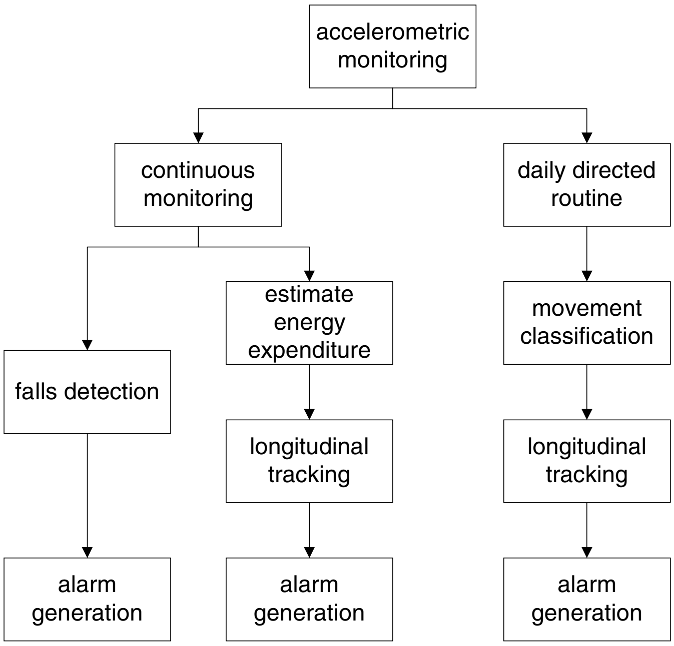

Figure 6. The three modes of monitoring.

continuous transmission. The wearable unit also contained a push button so that it could be used as a manual personal alarm system (Salleh *et al* 2000).

A two-stage approach to monitoring was adopted. Each morning, subjects were required to carry out a short routine of important activities, namely, standing, sitting into a chair, rising from a chair, walking, lying down, and standing up from lying. All of these movements are basic to independent living at home. Parameters including step rate, step rate variability, rise time, postural orientation during sitting and standing, postural sway during quiet standing and the energy expended in each movement were extracted from the signals. Subjects then wore the monitor for the rest of the day, during which period they were continuously monitored for adverse events such as falls, and parameters including hourly and daily activity levels and energy expenditure, and the amount of time spent each day in activity and in rest were tracked longitudinally. Figure 6 shows the functions that were performed by the system.

A recent pilot study, in which the system was used daily by six healthy elderly subjects over a 2–3 month period, demonstrated the feasibility of this approach. High levels of compliance were achieved and all of the subjects reported that the system was easy to use and the monitor comfortable and unobtrusive to wear. Parameters of movement were successfully extracted from the signal and able to be tracked longitudinally (Mathie *et al* 2003a, 2003c). The next stages of this work are to use the integrated monitoring system with frail or ill, housebound patients in order to assess the clinical utility of longitudinal monitoring of parameters for this cohort and to monitor elderly fallers with the system to validate the falls detection component of the system for monitoring of free-living subjects.

The importance of integrated, ambulatory monitoring has been acknowledged in the context of other ambulatory monitoring systems (Asada *et al* 2003, Park and Jayaraman 2003, Jovanov *et al* 2003, Winters *et al* 2003). The feasibility of assessing individual movements using accelerometers has been demonstrated, and preliminary research such as that described above has shown their potential in an integrated system. However, the full potential of accelerometry for integrated monitoring has yet to be fully explored.

7. Conclusion

Accelerometry is a tool that is suitable for long-term monitoring of free-living subjects because it can provide objective, reliable monitoring of unconstrained subjects for low cost. A wide range of measures, including classification of movements, assessment of physical activity level, estimation of metabolic energy expenditure, assessment of balance, gait and sit-to-stand transfers can be reliably obtained. Many of these functions can actually be carried out using a single triaxial accelerometer worn at the waist.

There are several modes in which an accelerometry system may be used. It may be used for one-time clinical assessment. Alternatively, it may be used for long term, unsupervised monitoring for adverse event detection, and for longitudinal tracking of relevant physiological parameters. The different functions and modes of operation of an accelerometry monitoring system can be integrated to provide a more comprehensive, intelligent home monitoring system that provides additional security to people living alone at home, and also provides valuable clinical information on functional ability in the home environment. The longitudinal monitoring of parameters has the potential to provide valuable information on ongoing health status and functional ability during daily activity.

Studies and investigations so far have shown that accelerometers are a valuable tool for research purposes as well as for monitoring human movements and can be utilized both in conjunction with other techniques and as stand-alone devices. Further integration and development of techniques for these devices will widen their scope and applicability in the study of human movement.

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