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A Triaxial Accelerometer and Portable Data Processing Unit for the Assessment of Daily Physical Activity

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***Abstract*— The present study describes the development of a triaxial accelerometer (TA) and a portable data processing unit for the assessment of daily physical activity. The TA is composed of three orthogonally mounted uniaxial piezoresistive accelerometers and can be used to register accelerations covering the amplitude and frequency ranges of human body acceleration. Interinstrument and test–retest experiments showed that the offset and the sensitivity of the TA were equal for each measurement direction and remained constant on two measurement days. Transverse sensitivity was significantly different for each measurement direction, but did not influence accelerometer output (<3% of the sensitivity along the main axis). The data unit enables the on-line processing of accelerometer output to a reliable estimator of physical activity over eight-day periods. Preliminary evaluation of the system in 13 male subjects during standardized activities in the laboratory demonstrated a significant relationship between accelerometer output and energy expenditure due to physical activity, the standard reference for physical activity ($r = 0.89$). Shortcomings of the system are its low sensitivity to sedentary activities and the inability to register static exercise. The validity of the system for the assessment of normal daily physical activity and specific activities outside the laboratory should be studied in free-living subjects.**

***Index Terms*—Assessment of daily physical activity, data processing unit, energy expenditure, physical activity, piezoresistive accelerometer, portable unit, triaxial accelerometer**

I. Introduction

THE quantitative assessment of daily physical activity in humans requires an objective and reliable technique to be used under free-living conditions. From a physiological point of view, physical activity can be regarded as any movement or posture that is produced by skeletal muscles and results in energy expenditure [1]. Currently, the energy expenditure due to physical activity is widely accepted as the standard reference for physical activity [2], but measurement of this variable under conditions of daily living is impractical and not feasible for population studies. Therefore, the interest for estimates of energy expenditure based on observations, questionnaires, heart rate recordings, or movement registration is growing. At present, movement registration with body-fixed motion sensors offers the best alternative for daily physical activity assessment. Various motion sensors have been designed for this purpose, ranging from mechanical pedometers [3] and actometers [4] to electronic accelerometers [5]–[7]. Unfortunately, motion sensors cannot be used to determine the static characteristics of physical activity. It is assumed, however, that the contribution of static exercise to the total level of physical activity is negligible under normal daily living conditions [8], [9].

Electronic accelerometers are the most promising motion sensors for physical activity assessment in free-living subjects. These sensors respond to both frequency and intensity of movement, and in this way are superior to pedometers and actometers, which are attenuated by impact or tilt and only count body movement if a certain threshold is passed. Due to the current state of art in integrated circuit technology there is also good opportunity to build very small and lightweight accelerometer systems that can be worn for days or even weeks. Accelerometers have been used for several decades to study gait and other movements [10], [11] or for the measurement of tremor and motor activity in neurological patients [12], [13]. The use of accelerometers for the assessment of physical activity is based on demonstrated relationships between accelerometer output and energy expenditure in studies on gait analysis and ergonomics. In studying the vertical forces resulting from body movement in industrial workers with a force platform, Brouha [14] found a significant correlation between the integral of the rectified force-time curve and energy expenditure ($r = 0.83$–0.96). Ismail *et al*. [15] measured forces during walking in three orthogonal planes by using a force platform placed under the belt of a treadmill. By integration of the rectified force-time curves it was possible to predict energy expenditure from each measurement direction ($r = 0.73$–0.92). The best prediction of energy expenditure, however, was achieved by summation of the absolute values of the three orthogonal forces. Reswick *et al*. [16] used a head-mounted accelerometer during walking on a large walkway. They concluded that the integral of the modulus of accelerometer output was linearly related to the energy expenditure during walking. These results have led several researchers to hypothesize that the integral of the modulus of acceleration measured on the human body—especially in vertical direction—can be used to predict energy expenditure due to physical activity. In 1981, Wong *et al*. [7] developed a portable motion sensor with a piezo-electric accelerometer

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to estimate energy expenditure during various activities. This sensor, named Caltrac, could be attached to the waist and registered accelerations parallel to the vertical axis of the body. The absolute value of accelerometer output was integrated and summed for the total time it was worn. This measure was referred to as "acceleration count." Reproducibility of acceleration counts measured at the low back during various exercises was good ($r = 0.94$) and a pooled correlation coefficient of 0.74 was found for the relationship between acceleration count and energy expenditure [6]. From more recent studies it can be concluded that Caltrac output shows good correlation with measured energy expenditure during separate well-defined activities in the laboratory, like treadmill walking ($r = 0.68$–0.94), but generally overestimates energy expenditure under these circumstances [17], [18]. Under free-living conditions the uniaxial Caltrac can be used to distinguish among interindividual levels of daily physical activity, but tends to systematically underestimate the energy requirements when compared to whole body indirect calorimetry [19] or the doubly labeled water method [20], [21]. It seems obvious that the use of the uniaxial Caltrac limits the registration of multidirectional body movement and hence the prediction of energy expenditure. This was demonstrated by Ayen and Montoye [22] who showed that energy expenditure during exercises like walking, running, and squat-thrusts is better predicted by using three separate Caltracs, mounted at right angles on the waist ($r = 0.75$), than by using a single vertical Caltrac ($r = 0.65$). Recently, Meijer *et al*. [5] developed a portable triaxial accelerometer (TA) for the assessment of daily physical activity. This accelerometer was based upon a bended piezo-electric plate, sensitive to accelerations in three directions. A data-acquisition unit with solid-state memory was used for lowpass filtering, amplification, rectification, and integration of the resulting accelerometer output over 1-min time intervals. In the laboratory a linear relationship between accelerometer output and energy expenditure was found for the pooled data of 16 subjects performing specified exercises resembling activities of daily living [23]. However, at high levels of energy expenditure, especially in running, the accelerometer output systematically underestimated energy expenditure. During evaluations under free-living conditions over 50% of the measured data were lost due to bad performance of the devices [24]. Other shortcomings of the device used by Meijer *et al*. are the higher sensitivity in vertical measurement direction compared to the horizontal measurement directions, the lack of information from separate measurement directions, and the lack of a dc response for adequate calibration of the accelerometer. This last item is manifest in all piezo-electric accelerometers.

The primary aim of the present study was to develop an improved electronic TA device for daily physical activity assessment. Initial investigation was aimed at specifying the type of accelerometer to be used. For this purpose, the frequency and amplitude characteristics of human body acceleration, which will ultimately determine the technical specifications of the accelerometer, were described. Next, a TA was designed. In order to investigate the relative contribution of different measurement directions to the assessment of physical activity, this TA was composed of three separate, orthogonally mounted uniaxial accelerometers. The uniaxial sensors in the TA were evaluated concerning their interinstrument and test–retest variability in a bench-test. Finally, a portable data unit for the on-line acquisition, processing, and storage of accelerometer output over prolonged periods of time was developed to study patterns of daily physical activity. Preliminary evaluation of the TA and the data unit against measurements of energy expenditure due to physical activity (EE_{act}) was performed during a standardized activity protocol in a respiration chamber.

II. Development of a Triaxial Accelerometer

When choosing an appropriate accelerometer for the assessment of daily physical activity one should consider the specifications of the various available electronic accelerometers as well as the characteristics of human movement which determine the output of a body-fixed accelerometer. These aspects are described below.

A. Electronic Accelerometers

Electronic accelerometers are based on piezo-electric or piezoresistive properties. Piezo-electric accelerometers can be considered as damped mass-spring systems, in which a piezo-electric element acts as spring and damper. This element generates an electrical charge in response to the mechanical force, and hence, the acceleration applied to it by a small mass. In piezoresistive accelerometers "spring and damper" are replaced by silicon resistors which change electrical resistance in response to the applied mechanical load. The resistors are electrically connected in a Wheatstone bridge to produce a voltage proportional to the amplitude and frequency of the acceleration of the small mass in the sensor. Piezoresistive accelerometers are smaller than piezo-electric accelerometers, but require an external power source. Furthermore, piezoresistive accelerometers have dc response, while piezo-electric accelerometers do not respond to constant acceleration.

B. Sources of Accelerometer Output

The output of an ideal accelerometer worn on the human body originates from several sources: 1) acceleration due to body movement; 2) gravitational acceleration; 3) external vibrations, not produced by the body itself (e.g., resulting from vehicles); 4) accelerations due to bouncing of the sensor against other objects or jolting of the sensor on the body due to loose attachment, eventually resulting in mechanical resonance [25]. Of these, only the first two are directly related to intentional movement of the body. Gravitational acceleration may vary between -1 and 1 g, depending on the orientation of the measurement direction of the sensor in the gravitational field. This source of accelerometer output is often described as the "gravitational component." The output due to body movement, usually referred to as the "kinematic component," is dependent on the type of activity performed, the part of the body where accelerations are measured, and the measurement direction (antero-posterior, medio-lateral, or vertical), as is discussed below. Sources 3 and 4, which may add "noise" to the accelerometer output, can be attenuated by adequate filtering techniques (if the frequency range of the noise does not interfere with the frequency range of human body acceleration) and proper attachment of the sensor to the body.

C. Human Body Acceleration

Accelerometers used for the assessment of physical activity in humans must provide accurate registrations of the frequencies and amplitudes of accelerations involved in human movement. Once the broadest possible spectra of frequencies and amplitudes is known, the right sensor can be chosen. In general, frequencies and amplitudes of accelerations involved in human movement are relatively low. The largest accelerations with the highest frequencies might be expected during running and jumping.

1) Frequency: During locomotion, frequencies are generally higher in the vertical than in the medio-lateral or the antero-posterior directions and the frequency spectrum shifts toward higher frequencies from cranial to more caudal parts of the body. In walking at natural velocity the bulk of acceleration power in the upper body ranges from 0.8–5 Hz, whereas the most abrupt accelerations occur at the foot in vertical direction during heel strike and sometimes amount up to 60 Hz [26]. By measuring these “worst case” accelerations at heel strike with a force platform Antonnson and Mann [27] demonstrated that 99% of the acceleration power during walking with bare feet is concentrated below 15 Hz. Higher frequencies are caused by the impact between foot and walking surface and do not directly result from voluntary muscular work. By using piezoresistive accelerometers (range: ± 20 g, frequency response: 0–70 Hz), taped to the skin, Bhattacharya *et al*. [28] found the majority of frequency components during running to vary between 1–18 Hz in vertical direction at the ankle. At the low back and the head, the frequency content of acceleration profiles was smaller. During trampoline jumping, where the impact between foot and “jumping surface” is less pronounced than in running, the frequency content of vertical accelerations at the ankle, the low back, and the head was approximately equal, ranging from 0.7–4 Hz. The frequency content of daily activities has recently been studied by Sun and Hill [29]. Fast Fourier analysis of daily activities performed on a force platform revealed the major energy band to be between 0.3–3.5 Hz.

2) Amplitude: Like the frequency characteristics, the amplitudes of accelerations involved in locomotion are usually higher in the vertical direction than in both horizontal directions and increase in magnitude from cranial toward caudal body parts. During walking, for instance, the accelerations of the upper body determined from stereophotogrammetry range from -0.3 to 0.8 g in the vertical direction, whereas in the horizontal directions they range from -0.2 to 0.2 g at the head and from -0.3 to 0.4 g at the low back [26]. At the tibia, the amplitude of accelerations during walking varies between -1.7 and 3.3 g in the vertical direction and between -2.1 and 2.3 g in the horizontal directions, as measured with bone-mounted piezoresistive accelerometers (range: ± 25 g, f_0: 1000 Hz) [30]. During running, Bhattacharya *et al*. [28] observed absolute vertical peak accelerations ranging from 0.8–4.0 g at the head, from 0.9–5.0 g at the low back, and from 3.0–12.0 g at the ankle by using their skin-mounted accelerometers. During trampoline jumping, differences in absolute vertical peak accelerations measured at the head, the low back, and the ankle were less pronounced than in running: they varied between 3.0–5.6 g at the head, between 3.9–6.0 g at the low back, and between 3.0–7.0 g at the ankle.

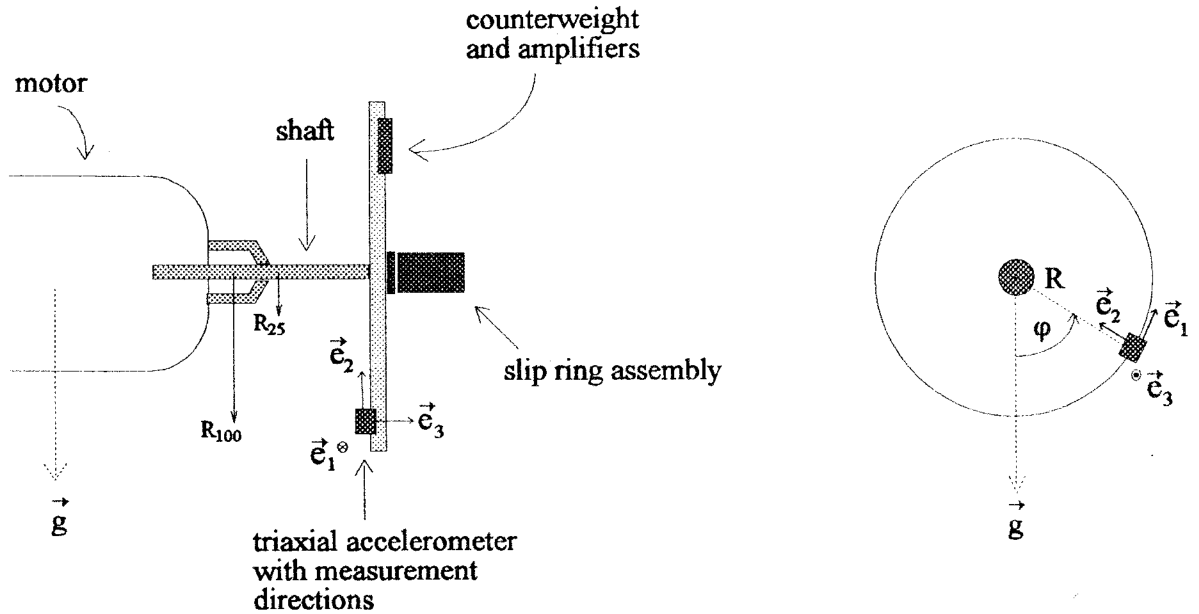
3) Desirable Frequency and Amplitude Ranges: Despite the difference in measurement techniques used to determine the frequency and amplitude spectra of human body accelerations, the above-mentioned studies do provide insight into the frequencies and amplitudes that might be expected during normal daily activities. Considering the findings of these studies, body-fixed accelerometers must be able to register accelerations within the amplitude range of -12 to $+12$ g and with frequencies up to 20 Hz in order to assess daily physical activity. In general, body-fixed accelerometers for physical activity assessment are placed at waist level [5], [23], [31]. At this site an amplitude range of about -6 to $+6$ g will suffice. Although the major kinematic acceleration component during human movement is usually found in the vertical direction, accelerations in other directions do contribute to the total, complex movement pattern. Thus, for a complete registration of multidirectional human movement, accelerations should be measured in three directions. However, to investigate the relative contribution of separate measurement directions to the estimation of EE_{act}, it should be possible to analyze the output of the three measurement directions independently.

D. The Triaxial Accelerometer

An accurate registration of the frequencies and amplitudes of accelerations involved in human movement is dependent on the type of accelerometer used. As is mentioned above, piezoresistive accelerometers yield a dc response, whereas piezo-electric accelerometers do not. A major advantage of dc response is that it enables the calibration of piezoresistive accelerometers, for instance by rotation within the gravitational field. Human movement, however, will never correspond to a dc response. Therefore, it was decided to use a piezoresistive sensor in combination with a highpass filter to eliminate any resulting dc component during the registration of human movement, while retaining a very low frequency cutoff (about 0.1 Hz). A number of different uniaxial piezoresistive accelerometers was investigated and a small, lightweight accelerometer (ICSensors, type 3031-010, size: $4 \times 4 \times 3$ mm; weight 0.3 g; range: ± 10 g, frequency response: 0–600 Hz; f_0: 1200 Hz) was selected for further research. Three uniaxial accelerometers (A_1, A_2, and A_3) were mounted orthogonally onto a $12 \times 12 \times 12$-mm cube made of Celeron to accomplish a TA with independent measurement directions.

III. Evaluation of the Triaxial Accelerometer

In order to assess the usefulness of the TA for the measurement of accelerations involved in human movement, the device was tested in a laboratory experiment. The output of the three uniaxial accelerometers A_1, A_2, and A_3 as a consequence of a mechanically applied acceleration was studied under various conditions. Interinstrument and test–retest comparisons were



(a) (b)

Fig. 1. Experimental system to evaluate interinstrument and test–retest variability of accelerometer characteristics. (a) Lateral view of equipment ($\phi = 0$). (b) Frontal view of the rotational motion applied to a TA ($\phi \neq 0$). System-fixed measurement directions ($\vec{e}_1$, $\vec{e}_2$, $\vec{e}_3$) are indicated. Radius R represents the distance between the TA and the center of rotation, $\vec{g}$ is the gravitational acceleration vector, and ϕ the angle between R and $\vec{g}$. R_{25} and R_{100} refer to radii of 25 and 100 mm, respectively.

made to test whether the outputs of A_1, A_2, and A_3 were similar under the same experimental conditions and remained constant on different occasions.

A. Methods

A schematic representation of the experimental setup is given in Fig. 1. The TA was mounted onto a counterweighted arm at distances of 25 mm (R_{25}) or 100 mm (R_{100}) from the central shaft. After mounting, the positive measurement directions of the TA were parallel to the axes of the local system of reference ($\vec{e}_1$, $\vec{e}_2$, $\vec{e}_3$). By turning the TA around 90° angles the uniaxial accelerometers could be tested in different directions. Bridge amplifiers (PMI, type: OP9OGS) for the accelerometers were attached to the opposite side of the counterweighted arm. Connections between the amplifiers and the TA were established via a 12-conductor shielded cable. The central shaft of the arm was secured to the electric motor of a lathe (Laagland Rotterdam, type: Celtic 14) with an adjustable number of revolutions per second (rps) to cause a rotational motion of the arm with constant angular velocity (accuracy <0.1 rps). A slip ring assembly (Hottinger Baldwin Messtechnik, type: SK12) was used for the transmission of the output voltages of the three separate accelerometers to a computer. The supply voltage to the accelerometers and amplifiers was also provided by means of the slip ring assembly.

The rotational movement of the arm resulted in a constant radial acceleration in the $\vec{e}_2$ direction and gravitational components along $\vec{e}_1$ and $\vec{e}_2$, varying sinusoidally between -1 and 1 g (g= 9.812 m $\cdot$ s^{-2} at the experimental site) with a basic frequency equal to the applied number of rps. No accelerations were applied in $\vec{e}_3$ direction. Apart from the radial (kinematic) and gravitational components, the accelerometer output during the experiments resulted from the overall offset of the accelerometer and the experimental system, and—since the accelerometers were assumed to be nonideal—from transverse sensitivity. This last component is defined as the sensitivity of the accelerometers to accelerations from directions other than the measurement direction. Thus, accelerometer outputs a_1 and a_2 in $\vec{e}_1$ and $\vec{e}_2$ direction, including offset and transverse sensitivity, can be described by

$$a_{1,i} = V_{0,i} + G_i[-g\sin(\phi)] + k_i[R\dot{\phi}^2 - g\cos(\phi)] \quad (1)$$

$$a_{2,i} = V_{0,i} + G_i[R\dot{\phi}^2 - g\cos(\phi)] + k_i[-g\sin(\phi)] \quad (2)$$

with the index i ($i = 1, 2, 3$) denoting the number of the uniaxial accelerometer tested in the defined measurement direction. $V_{0,i}$ represents the offset, G_i the sensitivity along the main axis, and k_i the transverse sensitivity of the ith accelerometer. ϕ represents the angle between radius R and the gravitational vector of the earth ($\vec{g}$), $\dot{\phi}$ the time derivative of ϕ, and $R\dot{\phi}^2$ the radial acceleration. Accelerometer output is measured in mV, G_i is expressed as mV $\cdot$ g^{-1}, and k_i is expressed as mV $\cdot$ g^{-1} or as a percentage of G_i (%G_i). Note that along $\vec{e}_1$ and $\vec{e}_2$ the transverse sensitivity due to output from the $\vec{e}_3$ direction was neglected since no accelerations were applied in $\vec{e}_3$ direction. Accelerometer output along $\vec{e}_3$ is only produced by offset and transverse sensitivity due to accelerations in $\vec{e}_1$ and $\vec{e}_2$ direction

$$a_{3,i} = V_{0,i} + k_i[-g\sin(\phi)] + k_i'[R\dot{\phi}^2 - g\cos(\phi)] \quad (3)$$

TABLE I
OFFSET ($V_{0,i}$), SENSITIVITY (G_i), AND TRANSVERSE SENSITIVITY (k_i, GIVEN AS ABSOLUTE VALUE AND AS A PERCENTAGE OF G_i) FOR EACH OF THE THREE UNIAXIAL ACCELEROMETERS (A_1, A_2, AND A_3) IN THE TRIAXIAL ACCELEROMETER. VALUES ARE GIVEN AS MEAN ±SD FROM 33 OBSERVATIONS

accelerometer	$V_{0,i}$ (mV)	G_i (mV·g^{-1})	k_i (mV·g^{-1})	k_i (% G_i)
A_1	186 ± 47	581 ± 17	6.9 ± 0.3*	1.2 ± 0.4#
A_2	180 ± 31	575 ± 9	3.0 ± 0.2*	0.5 ± 0.1#
A_3	186 ± 51	577 ± 13	13.8 ± 9.9*	2.6 ± 1.7#

*: $p < 0.001$, significant difference between absolute values for k_i.
#: $p < 0.001$, significant differences between k_i (% G_i).

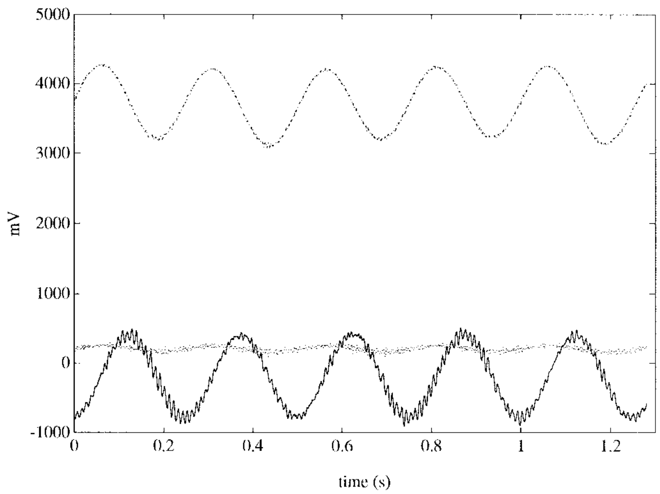


Fig. 2. Typical example of TA output in time in $\vec{e}_1$, (—), $\vec{e}_2$ (--), and $\vec{e}_3$ (···) direction during rotational motion at 4 rps with a radius of 100 mm.

where k_i, the transverse sensitivity due to acceleration along $\vec{e}_1$, and k_i', the transverse sensitivity due to acceleration along $\vec{e}_2$, are assumed to be equal.

By using 11 combinations of 1–10 rps with the two radii, R_{25} and R_{100}, the constant acceleration along $\vec{e}_2$ was varied from 0.1–15.6 g (1.2–152.7 $\text{m} \cdot \text{s}^{-2}$). Accelerometer outputs during combinations of R_{25} with rps >10 and R_{100} with rps >6 could not be measured, since the output voltage exceeded the maximal allowed output voltage of the amplifiers (±15 V). The total amplitude range of human body acceleration, however, was included in the experiments. For each trial the outputs from A_1, A_2, and A_3 were sampled and analyzed synchronously using a data-acquisition board (DIFA Measuring Systems). Signals were sampled at 800 Hz to produce a representation of accelerometer output in time. A typical example of accelerometer output in time along $\vec{e}_1$, $\vec{e}_2$, and $\vec{e}_3$ is shown in Fig. 2 for rotation at 4 rps at a radius of 100 mm. Next, the amplitude spectrum of the signals was determined. The average spectrum of four subsequent measurements of 2048 samples, sampled at 25 Hz, was used for further calculations. The frequency of the measured accelerations could be determined with an accuracy of 0.01 Hz.

The dc components of the amplitude spectra were used to determine $V_{0,i}$ plus the output due to constant radial acceleration in $\vec{e}_2$ direction. The difference between accelerometer outputs from trials performed at the same angular velocity but at different radii was taken to eliminate $V_{0,i}$ and to determine the sensitivity due to the radial acceleration. In the same way the transverse sensitivity due to the radial accelerations in $\vec{e}_1$ and $\vec{e}_3$ direction were determined. Frequency components of the amplitude spectra were windowed with a flat-top window before they were used to compute the sensitivity and transverse sensitivity due to gravitational acceleration in $\vec{e}_1$ and $\vec{e}_2$ direction and the transverse sensitivity due to gravitational acceleration in $\vec{e}_3$ direction. Besides the determination of $V_{0,i}$, G_i, and k_i, it was investigated whether accelerometer output along $\vec{e}_2$ was proportional to the applied radial acceleration within the amplitude range of human body acceleration.

The experimental protocol of 11 trials was repeated three times on day one, with the TA positioned in another orientation on each repetition. Thus, 33 combinations of R, rps, and orientation of the TA were performed to calculate $V_{0,i}$, G_i, and k_i under different conditions. Data are presented as mean and standard deviation (sd). Interinstrument variability in $V_{0,i}$, G_i, and k_i was statistically analyzed with a repeated-measures analysis of variance (ANOVA), followed by a post-hoc test (Scheffe F-test) to indicate possible significant differences. A level of 5% ($p \leq 0.05$) was taken as level of significance. Part of the experimental protocol was performed on two separate occasions (days one and two) to determine test–retest variability in $V_{0,i}$, G_i, and k_i. Only trials at 2, 4, and 6 rps were included is this analysis. Statistical comparisons between the measured parameters on days one and two were made by using a paired T-test.

Accelerations with frequencies above 10 Hz were not included in the experiments. It was assumed, however, that frequencies within the range of human body movement (up to 20 Hz) could be measured accurately with the uniaxial accelerometers which have a frequency response of 0–600 Hz. To verify this assumption, one of the uniaxial accelerometers (A_1) was tested using a vibration excitator (Ling Dynamic Systems, type: 201) at frequencies between 0–35 Hz and amplitudes of 0.50–1.25 mm. The amplification of accelerometer output was similar to that during the interinstrument and test–retest experiments.

B. Results

Values for $V_{0,i}$, G_i, and k_i on day one are given in Table I. $V_{0,i}$ and G_i were similar for A_1, A_2, and A_3, whereas the absolute values of k_i, as well as k_i expressed as a percentage of G_i, were different for A_1, A_2, and A_3. In all three accelerometers, however, k_i did not significantly influence accelerometer output (<5% of the sensitivity along the main axis) and did not exceed the maximal value for k_i given by the manufacturer (3% G_i). No differences in parameter values were found for the separate measurement directions. For instance, the transverse sensitivity of accelerometer A_1, was similar when A_1, was tested along $\vec{e}_1$, $\vec{e}_2$, or $\vec{e}_3$.

Fig. 3 shows the dc components of accelerometer output resulting from constant angular velocity along $\vec{e}_2$ for accelerometers A_1, A_2, and A_3 plotted against the applied constant radial acceleration. Accelerometer output was corrected for offset values. The figures show that within the amplitude range of human body acceleration ($\pm$12 g), the dc component of accelerometer output is linearly related to the applied radial acceleration. The sensitivity of the accelerometers, as determined from the slope of the regression lines in Fig. 3, was 583 mV $\cdot$ g $^{-1}$ in A_1, 573 mV $\cdot$ g^{-1} in A_2, and 578 mV $\cdot$ g $^{-1}$ in A_3.

Values for $V_{0,i}$, G_i, and k_i (mean $\pm$ sd) on days one and two are indicated in Table II. Note that average values on day one may differ from those in Table I, since only 18 observations were used for test–retest analysis, while 33 observations were included in the interinstrument analysis.

No differences in parameter values were found between days one and two. Values for k_i, either expresses as mV $\cdot$ g^{-1} or as %G_i, were significantly different from each other on both days. Again, in all three accelerometers k_i was well below 5% of G_i. Accelerometer output as a function of the applied radial acceleration along $\vec{e}_2$ on day two was comparable to the data given in Fig. 3.

During vibrational motion with frequencies up to 35 Hz the amplitude of the output from A_1 (average of three trials, corrected for offset) was linearly related to the amplitude of the applied acceleration (Fig. 4). The sensitivity, G_1 of A_1, during the vibration experiments, as determined from the slope of the regression line, was 580 mV $\cdot$ g $^{-1}$.

From these results it was concluded that the uniaxial accelerometers of the TA could be used for the registration of accelerations within the amplitude and frequency range of human body acceleration. The accelerometers were built in a flat housing of Celeron ($50 \times 30 \times 8$ mm, 16 gr) with two slits for an elastic belt. By using this belt the TA could easily be attached to several locations on the human body.

IV. Development of a Data Processing Unit

In order to correlate analog accelerometer output to discrete data on energy expenditure, the output must be processed to a useful quantity. Recently, Bouten *et al.* [32] reported on the most optimal way of data processing for the estimation of EE_{act} from accelerometer output. From synchronous recordings of EE_{act} and accelerometer output, measured with the TA attached to the low back, it was concluded that EE_{act} was better predicted from three-directional than from unidirectional accelerometer output when different types of activity (sedentary activities, walking) were performed. Summation of the time integrals of the moduli of accelerometer output from the separate measurement directions (IMA_{tot}), resulted

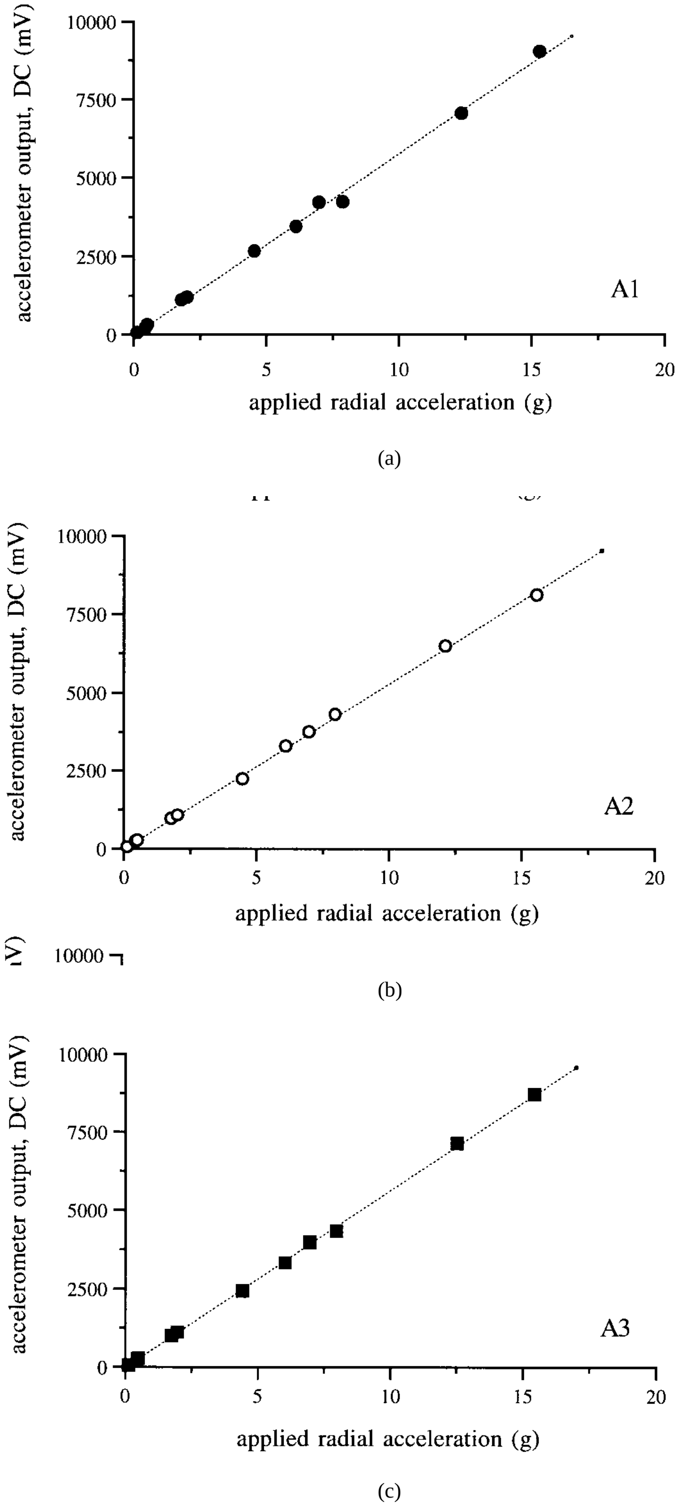


Fig. 3. Accelerometer output (dc-component), corrected for offset, as a function of the applied radial acceleration in three uniaxial accelerometers (A_1, A_2, and A_3).

in the most accurate prediction of EE_{act} ($r = 0.95$). The data processing referred to is given by

$$IMA_{\text{tot}} = \int_{t=t_0}^{t_0+T} |a_1|\, dt + \int_{t=t_0}^{t_0+T} |a_2|\, dt + \int_{t=t_0}^{t_0+T} |a_3|\, dt \tag{4}$$

TABLE II
OFFSET ($V_{0,i}$), SENSITIVITY (G_i), AND TRANSVERSE SENSITIVITY (k_i, GIVEN AS ABSOLUTE VALUE AND AS A PERCENTAGE OF G_i) OF THE THREE UNIAXIAL ACCELEROMETERS (A_1, A_2, A_3) IN THE TRIAXIAL ACCELEROMETER ON DAYS ONE AND TWO. VALUES ARE GIVEN AS MEAN ±SD FROM 18 OBSERVATIONS

		day 1	day 2
$V_{0,1}$	(mV)	181 ± 46	187 ± 44
$V_{0,2}$		178 ± 39	179 ± 32
$V_{0,3}$		180 ± 46	182 ± 40
G_1	($mV \cdot g^{-1}$)	589 ± 16	586 ± 11
G_2		587 ± 8	589 ± 7
G_3		584 ± 11	590 ± 13
k_1	($mV \cdot g^{-1}$)	6.9 ± 0.4*	6.9 ± 0.4*
k_2		3.0 ± 0.4*	3.0 ± 0.2*
k_3		4.7 ± 4.4*	4.6 ± 3.1*
k_1 (% G_1)	(%)	1.2 ± 0.3#	1.2 ± 0.4#
k_2 (% G_2)		0.5 ± 0.1#	0.5 ± 0.1#
k_3 (% G_3)		0.8 ± 0.6#	0.8 ± 0.6#

*: $p < 0.01$, significant differences between k_1, k_2, and k_3 on day one as well as on day two.
#: $p < 0.01$, significant differences between k_1 (% G_1), k_2 (% G_2), and k_3 (% G_3) on day one as well as on day two.

No significant differences in parameter values between days one and two were found.

with T the time period for integration and a_1, a_2, and a_3 the accelerometer outputs along the $\vec{e}_1$, $\vec{e}_2$, and $\vec{e}_3$ directions of a body-fixed system of reference. This data processing was superior to, for instance, the integration of the total acceleration vector.

When using IMA_{tot} for the assessment of physical activity in free-living subjects, the ambulatory recordings of this variable should not interfere with the subjects' physical activities. Therefore, a portable data unit with minimal weight and dimensions for the on-line acquisition, processing, and storage of accelerometer data is required. In addition, this data unit must enable the amplification and filtering of acceleration signals from the TA as well as the storage of IMA_{tot} over periods of days or weeks to study patterns of daily physical activity. Considering these demands, a free programmable data unit was developed for the assessment of daily physical activity with minimal discomfort to subjects.

A. The Data Processing Unit

The block diagram in Fig. 5 shows the processing of the output from the TA, which is implemented in the data unit. The connection between the data unit and the TA is established via a 0.5 m flexible 12 conductor shielded cable. Individual outputs from the three measurement directions of the TA are amplified and highpass (0.11 Hz, 5.6 dB/octave) and lowpass (20 Hz, 9 dB/octave) filtered to attenuate dc-response and frequencies that cannot be expected to arise from voluntary human movement. Next, acceleration signals are digitized (100 Hz) and further processed using a miniaturized datalogger (Onset Computer Corporation, type: Tattletale 5F). This datalogger enables a free programmable data processing and is programmed and started from a computer via a serial interface (Onset Computer Corporation, type: TC-1). For the assessment of daily physical activity the datalogger was programmed to calculate IMA_{tot}, (4), using its TX-Basic software package. The time period for integration is variable and can be adjusted at the start of a measurement period. After processing, the obtained data are stored in a 512 kB, 16 bit data memory chip that can be read out with the serial interface to a computer. There is also the possibility to start the datalogger and to read out the memory chip by modem. The memory is reset by disconnecting the supply voltage to the TA and the data unit, which is provided by batteries. Two 9 V, 1200 mAh batteries are required to register and process acceleration signals over a period of eight days. For a measurement period of three days two 9 V, 500 mAh batteries can be used. Batteries, datalogger, and other electronic components for data processing are built in a housing of PVC. This housing can be opened by the investigator for replacement of batteries and calibration of the accelerometers. The ac/dc switches within the housing are used to omit the high pass filters. The dc responses of the uniaxial accelerometers can than be applied to determine their sensitivity by altering the orientation of the TA with respect to the gravitational vector of the earth. The gain of the accelerometers as well as the balance of the Wheatstone bridge in the accelerometers (zeroing) can be adjusted manually. In this way the separate measurement directions of the TA can be calibrated equally. For the assessment of daily physical activity the gain is usually set to produce an accelerometer output of 1.5 V $\cdot$ g^{-1}, corresponding to an output count of 1000 over an integration period of 1 min in the data unit. The data unit measures $110 \times 70 \times 35$ mm and weighs 170 gr without batteries (250 gr including batteries). It can be worn around the waist in a small bag (fanny pack), but it can also be attached directly to a waist belt by using two slits on both sides of the PVC housing.

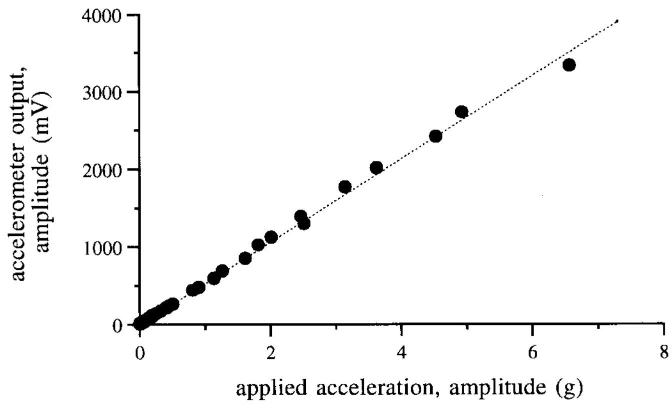


Fig. 4. Amplitude of measured accelerations, corrected for offset, as a function of the amplitude of the applied acceleration in a uniaxial accelerometer (A_1) during vibrational motion.

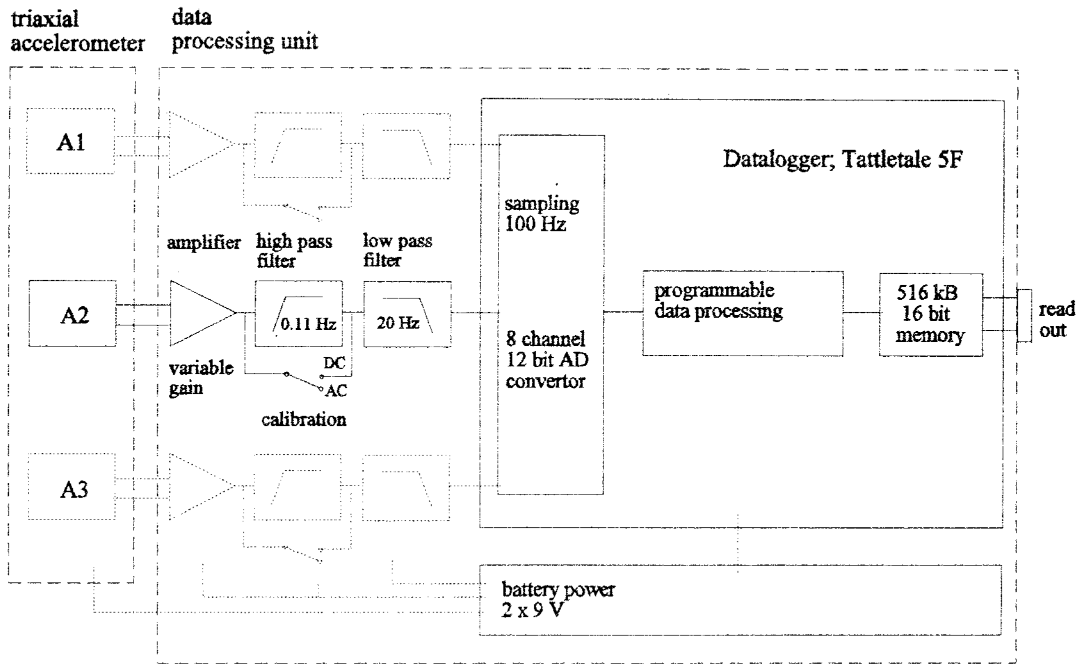


Fig. 5. Block-diagram of the TA and the data unit for acquisition, processing and storage of accelerometer output. The complete data processing is indicated for one of the uniaxial accelerometers (A_2).

B. Evaluation of the Triaxial Accelerometer and Data Processing Unit

Preliminary evaluation of the TA and data processing unit was performed in a group of 13 young male subjects (age: 27 ± 4 yr, body mass: 77 ± 12 kg, height: 1.83 ± 0.07 m) during a standardized long-term activity protocol in a respiration chamber [33]. This chamber (14 m^3) is furnished with a bed, table, chair, toilet, washing-bowl, radio, and television and is provided with equipment for the determination of metabolic energy expenditure from respiratory gas exchange. The ventilation rate through the chamber was measured with a dry-gas meter (Schlumberger, G4) and analyzed by a paramagnetic O_2 analyzer (Hartmann & Braun, Magnos 6G) and an infrared CO_2 analyzer (Hartmann & Braun, Uras 3G). Ingoing air was analyzed once every 15 min and outgoing air twice every 5 min. From the ventilation rate and O_2 and CO_2 concentrations in in- and out-going air, O_2 consumption and CO_2 production were calculated on-line on a computer. Total energy expenditure was calculated at 5-min intervals from O_2 consumption and CO_2 production, according to Weir [34]. The subjects stayed in the chamber for 36 h: two nights and the intervening day. During day-time (0830–2200) they performed standardized activities, resembling normal daily activities (sedentary activities, household activities, walking). Each activity was performed for 30 min. Except for stepping and carrying loads, the activities were performed at the subjects' preferred rate. Stepping was performed at 5-min intervals, alternately with 5-min rest periods, on a bench 33 cm high and at a rate of 30 steps $\cdot$ min^{-1} or 60 steps $\cdot$ min^{-1}. During carrying loads the subjects carried 1-kg iron disks from one side of the room to the other side at 5-min intervals. For each 30-min activity period the average total energy expenditure was determined. Next, EE_{act} for each activity was calculated as the average total energy expenditure minus sleeping metabolic rate, which was measured over a 3-h interval between 0230–0700 with the lowest level of activity as indicated by a Doppler radar system in the chamber. EE_{act} was expressed in watts and corrected for body mass (W $\cdot$ kg^{-1}). During the entire activity protocol the TA was attached to the low back of the subjects at the level of the second lumbar vertebra by using an elastic belt around the waist. The time interval for integration of accelerometer output was set to 1 min and IMA_{tot} (counts $\cdot$ min^{-1}) was averaged for each 30-min activity period. Associations between average IMA_{tot} and EE_{act} were determined with linear regression analysis, according to the least squares principle. Correlation coefficients (Pearson's r) and standard errors of estimate were calculated. Fig. 6 shows the means and standard deviations for IMA_{tot} and EE_{act} for the 13 subjects during the activity protocol. Individual correlations between IMA_{tot} and EE_{act} varied between 0.87–0.97 (mean: 0.92), whereas standard errors of estimate ranged from 0.2–0.5 W $\cdot$ kg^{-1} (mean: 0.3 W $\cdot$ kg^{-1}). When using the data of all subjects and all activities, a pooled correlation coefficient of 0.89 was found (pooled standard error of estimate: 0.4 W $\cdot$ kg^{-1}). Comparison between estimated and measured values of EE_{act} by using the pooled regression equation for IMA_{tot} versus EE_{act} resulted in an average overestimation of EE_{act} by 7.5% when the

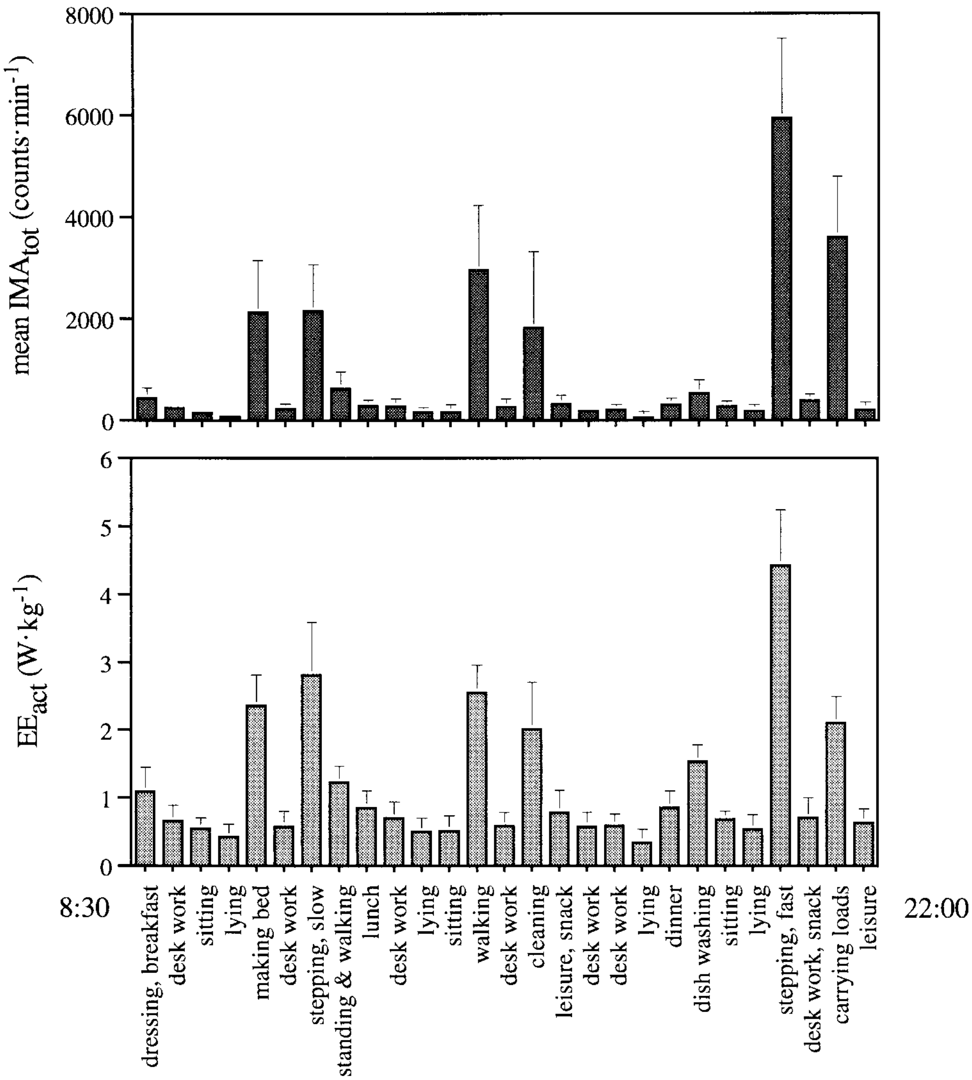


Fig. 6. Accelerometer output ($IMA_{\rm tot}$) and energy expenditure for physical activity ($EE_{\rm act}$) during a standardized activity protocol in a respiration chamber (mean $\pm$sd from 13 male subjects).

entire activity protocol was considered. Regarding the separate activities, differences between estimated and measured values of $EE_{\rm act}$ ranged from -40% (underestimation) during dish washing to $+33\%$ (overestimation) during lying. Intensive activities, like stepping, carrying loads, walking, cleaning, and making the bed, were underestimated on average by 6.2%, whereas less demanding activities, like sitting, lying, and desk work were overestimated on average by 6.6% using the accelerometer device.

V. Discussion

The present study describes the development and evaluation of a TA and data processing unit for the assessment of daily physical activity in terms of energy expenditure. The TA is composed of three separate orthogonally mounted uniaxial accelerometers in order to investigate whether $EE_{\rm act}$ is better predicted from three-directional than from unidirectional accelerometer output and to study the relative contribution of the separate uniaxial measurement directions to the estimation of $EE_{\rm act}$. These aspects could not be studied by the three-directional device described by Meijer *et al.* [5], [23] which consists of a single sensor, sensitive to multidirectional acceleration. Besides the lack of information from separate measurement directions, the device of Meijer *et al.* is also more sensitive to accelerations in the vertical direction than to accelerations in both horizontal directions. This might be a disadvantage when assessing activities like walking and running, since it has been shown that although the major acceleration component during these activities is in the vertical direction, $EE_{\rm act}$ is better predicted from accelerometer output in the antero-posterior direction [32]. As determined from interinstrument experiments, the sensitivity (G_i) of our TA was similar for each measurement direction—i.e., for each uniaxial accelerometer. No differences in offset values ($V_{0,\,i}$) were found, and the transverse sensitivity (k_i) did not influence accelerometer output. Test–retest experiments showed that values for G_i, $V_{0,\,i}$, and k_i were similar on two consecutive measurement days. Furthermore, it was concluded that, within the amplitude range of human body acceleration, the output of all three uniaxial accelerometers was proportional to the applied acceleration. For one of the accelerometers (A_1) it was shown that the output was proportional to the applied acceleration during vibrational motion at frequencies within the frequency range of human body acceleration. It is assumed that this is also true for A_2 and A_3. As can be observed from Tables I and II, relatively large deviations in k_3 were found. This is caused by the increase of k_3 with the magnitude of the applied acceleration. During accelerations above 10 g, k_3 reached values >5% G_3. However, as accelerations above 10 g are relatively scarce in human movement, it is assumed that k_3 will not affect the assessment of daily physical activity.

A major advantage of the piezoresistive accelerometers in the TA is their dc response, which facilitates calibration. Yet, the output voltage of these accelerometers is influenced by offset. In order to avoid over- or underestimation of the measured acceleration, accelerometer output must be corrected for offset values. The offset may drift, however, as a consequence of temperature changes, resulting in a more complicated correction for offset values. In our laboratory experiments the overall offset consisted of accelerometer offset and the offset of the amplifiers. Significant offset drift, typically 1% of the full scale over a temperature range of 0–50 °C in reference to 25 °C in the accelerometers and 1.2 $\mu V \cdot {}^{\circ}C^{-1}$ in the amplifiers, was not observed during these "short-term" experiments. During the "long-term" assessment of daily physical activity the offset—and possible offset drift—is attenuated by the highpass filter in the data unit. Nevertheless, offset and offset drift should be considered during calibration, when the highpass filter is omitted. The sensitivity of accelerometers may also drift due to temperature changes. Like the offset drift, the sensitivity drift of the piezoresistive accelerometers used in this study is typically 1% of the full scale and may affect the assessment of physical activity. However, no drift in sensitivity was observed during the laboratory experiments. From calibration data before and after the activity protocol in the respiration chamber, it was concluded that sensitivity and offset did also not drift during these long-term measurements.

Apart from body acceleration, the output of body-fixed accelerometers results from gravitational acceleration and noise due to external vibrations or inadequate attachment of the accelerometer. The gravitational component is dependent on the orientation of the accelerometer with respect to the gravitational vector of the earth. It may influence total accelerometer output considerably, especially when the angle ϕ between the measurement direction and the gravitational vector of the earth is relatively large and the kinematic component of accelerometer output is small. Consequently, it may affect the assessment of $IMA_{\rm tot}$ and hence the prediction of $EE_{\rm act}$. Correction for the gravitational component under daily living conditions is practically impossible. In order to minimize the effect of the gravitational component on accelerometer output, Servais *et al*. [31] argued that the attachment of accelerometers at locations where (the variation in) ϕ is small—e.g., the waist or the low back—is superior to locations where (the variation in) ϕ is large—e.g., the limbs. The precise effect of the gravitational component on $IMA_{\rm tot}$ and the relationship between $IMA_{\rm tot}$ and $EE_{\rm act}$, however, is unknown and should be studied.

External vibrations may also considerably influence $IMA_{\rm tot}$ under daily living conditions. A lowpass filter (20 Hz) is built into the data unit to attenuate frequencies that cannot be expected to arise from voluntary movement. Yet, contact of the accelerometer—or the subject wearing it—with vibrating external sources, like vehicles or machinery, may pose a major problem when frequencies of the external sources are interfering with the frequency range of human movement. For instance, the vibration of a power lawn mower (about 5.5 Hz) or the vibration of ground vehicles [25], [35] will affect accelerometer output and hence $IMA_{\rm tot}$.

The accelerometer should be fixed properly to the human body to avoid the sensor from moving or jolting on the skin. Attachment directly to the skin is essential, since movements of clothes will cause artifacts in accelerometer output. Preferably, the accelerometer is fixed with adhesive tape or elastic straps. In this way a firm attachment with minimal discomfort to subjects is achieved. Still, the soft tissue layer under the mounted accelerometer may affect accelerometer output, possibly leading to resonance of the accelerometer on the skin. Recently, Kitazaki and Griffin [36] studied the resonant behavior of skin-mounted accelerometers at the low back. Accelerometers with different masses and dimensions were attached to the skin over the spinous process of the third lumbar vertebra with double-sided adhesive tape. During a free vibration test in eight male subjects they found resonant frequencies as low as 15 Hz (range: 15–38 Hz) when using an accelerometer with similar measures as the TA (contact surface: 40×35 mm; total mass: 16 gr). Thus, like external vibrations with frequencies <20 Hz, these resonant frequencies may influence $IMA_{\rm tot}$ and should be considered during daily physical activity assessment.

The place of attachment of accelerometers on the human body is an important issue [37], [38]. First, the accelerometer on the human body may not interfere with the subjects' activities. Second, the kinematic and gravitational components of accelerometer output are dependent on the measurement location. We choose to attach the TA at the trunk (second lumbar vertebra) as this segment represents the major part of total body mass and is moving during most daily activities. Attachment of the TA to the low back caused minimal discomfort to the subjects and did not influence the performance of their activities. Furthermore, as discussed above, the effect of the gravitational component on total accelerometer output at this location is small. However, further research should provide evidence regarding the best place of attachment of accelerometers for physical activity assessment. In addition, the influence of small variations in place of attachment should be studied to test whether intersubject and intrasubject variability in placement affects the assessment of physical activity.

A portable data unit was developed for the on-line acquisition, processing, and storage of TA output. This data unit was programmed to calculate $IMA_{\rm tot}$, a variable that is highly related to the standard reference of physical activity: $EE_{\rm act}$ [32]. Evaluation of the TA and the data unit against $EE_{\rm act}$, during a standardized activity protocol resembling normal daily activity, showed strong individual ($r = 0.87$–0.97) and pooled ($r = 0.89$) correlations between $IMA_{\rm tot}$ and $EE_{\rm act}$. These correlations are higher than those found for standardized activities in the laboratory using the uniaxial Caltrac [6], [39]–[41] or the TA of Meijer *et al*. [5], [23]. Discrepancies between measured and estimated values of $EE_{\rm act}$ during high-intensity activities and low intensity activities were also smaller than those reported for the Caltrac [17]–[19] and the accelerometer of Meijer *et al*. [23].

At least part of the discrepancies between measured and estimated values of $EE_{\rm act}$ can be attributed to the performance of activities that involve static exercise, like stepping. During static exercise the increase in accelerometer output is not proportional to the increase in $EE_{\rm act}$ and the actual $EE_{\rm act}$ will be underestimated. Although it is assumed that the effects of static exercise on the total level of daily physical activity are negligible [8], [9], predictions of $EE_{\rm act}$ from accelerometer output during activities like walking uphill, carrying loads, or cycling with head wind should be evaluated with caution. The gravitational component of accelerometer output and noise from inadequate attachment of the TA may also have influenced the prediction of $EE_{\rm act}$ from $IMA_{\rm tot}$. In addition, it should be realized that the values of $EE_{\rm act}$ in this study are not corrected for the thermogenic effect of food consumption. Therefore, relatively high values of $EE_{\rm act}$, compared to $IMA_{\rm tot}$, may be found during activities following food consumption, as can be seen for dish washing. During this activity the actual $EE_{\rm act}$ was underestimated by 40% when using the pooled regression equation between $IMA_{\rm tot}$ and $EE_{\rm act}$.

It is obvious that the relatively high values for $IMA_{\rm tot}$ and $EE_{\rm act}$ during intensive activities may considerably influence the correlation between these variables. After elimination of $IMA_{\rm tot}$ values >1500 counts $\cdot$ min^{-1}, corresponding to making the bed, stepping, cleaning, walking, and carrying loads, individual correlations ranged from 0.57–0.83, with a mean of 0.80 (standard error of estimate: 0.1–0.3 W $\cdot$ kg^{-1}, mean: 0.1 W $\cdot$ kg^{-1}). The pooled correlation coefficient decreased to 0.77 (standard error of estimate: 0.3 W $\cdot$ kg^{-1}). The decrease in correlation is not only caused by the smaller ranges in $IMA_{\rm tot}$ and $EE_{\rm act}$ during less intensive activities, but also by the interindividual difference in performance of these activities. In addition, a relatively large contribution of the thermogenic effect of food consumption as well as a relatively large contribution of the gravitational component of accelerometer output may have affected the relationship between $IMA_{\rm tot}$ and $EE_{\rm act}$ during the less-intensive activities.

VI. CONCLUSION

The development of a TA and portable data unit has resulted in a new accelerometer device for physical activity assessment. Major advantages of the device are the use of piezoresistive accelerometers, which facilitates calibration, the ability to measure and analyze accelerations from three different measurement directions, and the on-line data processing to quantify accelerometer output as a function of physical activity ($IMA_{\rm tot}$). The calculation of $IMA_{\rm tot}$ over adjustable short-term intervals—for instance one min—enables the investigation of patterns of physical activity in time. From mechanical testing it was concluded that the TA was reliable and valid for the measurement of accelerations within the frequency and amplitude range of human body acceleration. Preliminary evaluation of the accelerometer device under laboratory conditions showed significant relationships between $IMA_{\rm tot}$ and $EE_{\rm act}$. These relationships, however, may be influenced by several factors, like the performance of static exercise, gravitational acceleration, and noise from external sources or resonant behavior of the accelerometer on the skin. These aspects, as well as the place of attachment of the TA, should be considered when evaluating the device in future applications. In addition, the validity and usefulness of the device for the assessment of $EE_{\rm act}$ under normal daily living conditions or during specific activities outside the laboratory should be studied in free-living subjects.

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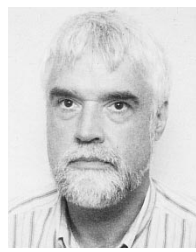

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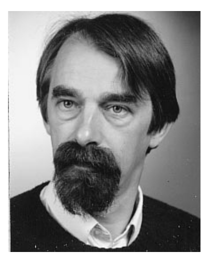

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He was appointed Professor of Continuums Mechanics at the same department in 1968. Since 1986, he has also been part-time Professor of Biomechanics at the Department of Health Sciences of the Maastricht University, The Netherlands. His main research interests lie in the areas of computational and experimental mechanics, biomechanics, and biomedical technology.