European Journal of Clinical Nutrition (2005) 59, 561–570

0954-3007/05 $30.00
www.nature.com/ejcn

ORIGINAL COMMUNICATION

Reliability and validity of the combined heart rate and movement sensor Actiheart

S Brage[1]*, N Brage[2], PW Franks[1], U Ekelund[1] and NJ Wareham[1]

[1]*MRC Epidemiology Unit, Institute of Public Health, University of Cambridge, UK; and* [2]*Institute of Sports Science and Clinical Biomechanics, University of Southern Denmark, Odense, Denmark*

Accurate quantification of physical activity energy expenditure is a key part of the effort to understand disorders of energy metabolism. The Actiheart, a combined heart rate (HR) and movement sensor, is designed to assess physical activity in populations.

Objective: To examine aspects of Actiheart reliability and validity in mechanical settings and during walking and running.

Methods: In eight Actiheart units, technical reliability (coefficients of variation, CV) and validity for movement were assessed with sinusoid accelerations (0.1–20 m/s^2) and for HR by simulated R-wave impulses (25–250 bpm). Agreement between Actiheart and ECG was determined during rest and treadmill locomotion (3.2–12.1 km/h). Walking and running intensity (in J/min/kg) was assessed with indirect calorimetry in 11 men and nine women (26–50 y, 20–29 kg/m^2) and modelled from movement, HR, and movement + HR by multiple linear regression, adjusting for sex.

Results: Median intrainstrument CV was 0.5 and 0.03% for movement and HR, respectively. Corresponding interinstrument CV values were 5.7 and 0.03% with some evidence of heteroscedasticity for movement. The linear relationship between movement and acceleration was strong ($R^2 = 0.99$, $P < 0.001$). Simulated R-waves were detected within 1 bpm from 30 to 250 bpm. The 95% limits of agreement between Actiheart and ECG were −4.2 to 4.3 bpm. Correlations with intensity were generally high ($R^2 > 0.84$, $P < 0.001$) but significantly highest when combining HR and movement (SEE < 1 MET).

Conclusions: The Actiheart is technically reliable and valid. Walking and running intensity may be estimated accurately but further studies are needed to assess validity in other activities and during free-living.

Sponsorship: The study received financial support from the Wellcome Trust and SB was supported by a scholarship from Unilever, UK.

European Journal of Clinical Nutrition (2005) **59,** 561–570. doi:10.1038/sj.ejcn.1602118
Published online 16 February 2005

Keywords: physical activity; heart rate variability; accelerometer; movement; walking; running; epidemiology; energy expenditure monitoring

Introduction

Physical activity is one of the most important environmental moderators of metabolism, and habitual physical inactivity was early identified as a major cause of metabolic diseases such as type II diabetes and heart disease (Paffenbarger *et al*, 1978; Saltin *et al*, 1979). However, physical activity is inherently difficult to measure precisely, especially when people are undergoing everyday activities. Therefore, it is unknown how much energy should be spent on activity to derive protection from metabolic disease (Kriska, 2000). Improving the assessment of energy expenditure would help clarify this dose–response relationship, enhance statistical power in association studies, enable monitoring of temporal trends, cross-cultural comparisons, and the effect of interventions, and finally allow better matching of energy intake to energy needs on the individual level (Wareham & Rennie, 1998; Alfonzo-Gonzalez *et al*, 2004). Added to the difficulty of measurement is the issue of the Hawthorne effect; behavioural modification caused by the act of being

***Correspondence: S Brage, MRC Epidemiology Unit , Institute of Public Health, University of Cambridge, Elsie Widdowson Laboratory, Fulbourn Road, Cambridge CB1 9NL, UK.**
E-mail: soren.brage@mrc-epid.cam.ac.uk
***Guarantors*: S Brage and NJ Wareham.**
***Contributors*: SB and NJW designed the study and SB was responsible for data collection. SB and NB conducted the data analysis, the results of which were discussed and interpreted by all authors. SB drafted the manuscript, which was reviewed by all authors.**
Received 11 February 2004; revised 27 August 2004; accepted 30 November 2004; published online 16 February 2005

observed (Wickstrom & Bendix, 2000; Tate *et al*, 2003). The extent to which this is a problem depends on the degree of invasiveness of the measurement method. Currently available methods have strengths and limitations with regard to precision, costs, practicality, and degree of invasiveness, and choice of instrument will depend on research objectives, evidence of reliability and validity at time of design, and available resources. Physical activity questionnaires are generally low cost, are relatively easily administered, and have low participant burden, but are poor at estimating energy expenditure on the individual level, most likely due to recall bias and coding errors (Albanes *et al*, 1990; Rennie & Wareham, 1998; Masse *et al*, 2002; Wareham *et al*, 2002; Ainslie *et al*, 2003). On the other hand, the doubly labelled water technique is relatively noninvasive and allows quantification of total energy expenditure over a prolonged period of time within 10% on the individual level and is thus considered the gold standard. The method is, however, expensive and cannot quantify subcomponents of physical activity energy expenditure (PAEE), such as intensity and frequency (Speakman, 1998; Ainslie *et al*, 2003). Modern motion sensors and heart rate (HR) monitors are capable of storing information about physical activity intensity (PAI) with high time resolution but the validity of such is compromised by different factors. Motion sensors display highly variable relationships to PAI in different activities (Montoye *et al*, 1983; Sekioka *et al*, 1998; Jakicic *et al*, 1999; Nichols *et al*, 1999; Bassett *et al*, 2000; Hendelman *et al*, 2000; Brage *et al*, 2003b, c; Ekelund *et al*, 2003; Strath *et al*, 2003; King *et al*, 2004; Treuth *et al*, 2004) whereas the main limitation with using HR is that it is affected by factors other than physical activity and the method relies more heavily on individual calibration (Andrews, 1971; Ceesay *et al*, 1989; Livingstone *et al*, 1990; Li *et al*, 1993; Strath *et al*, 2000). Thus, it remains a continuing goal to develop and evaluate methods for measuring PAEE and its subcomponents that are also affordable and minimally invasive. Methods which integrate physiological and motion detection systems have been identified as a promising research area (LaMonte & Ainsworth, 2001) and previous studies have demonstrated the utility of combined HR and movement measurement to improve precision of PAEE and PAI (Avons *et al*, 1988; Haskell *et al*, 1993; Moon & Butte, 1996; Luke *et al*, 1997; Treuth *et al*, 1998; Rennie *et al*, 2000; Strath *et al*, 2001; Strath *et al*, 2002; Brage *et al*, 2003b, c; 2004). However, no single commercially available device integrating HR and movement sensing has been described. The Actiheart (Cambridge Neurotechnology Ltd, Papworth, UK) is a single-piece combined HR and movement monitor, designed to clip on to two standard ECG electrodes on the chest (Figure 1). Before this device can be used for research purposes, however, aspects of reliability and validity must be evaluated. Therefore, the aims of the present study were to examine (a) the intra- and interinstrument reliability and validity of the HR and movement sensors within the Actiheart; (b) the monitor response in walking and running; and (c) the precision of preliminary prediction equations for walking and running intensity. We also investigated the utility of a step test to account for some of the variation in the individual HR–PAI relationships. This simple test may be used in field-based studies, where establishment of the individual HR–PAI relationships as measured by indirect calorimetry is not feasible.

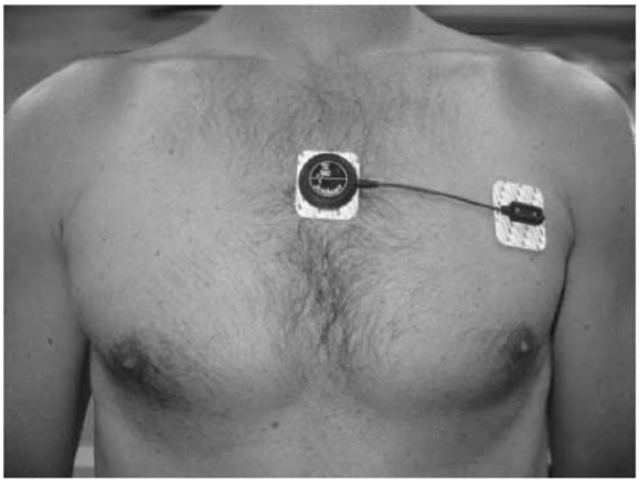

Figure 1 The Actiheart placed on human subject.

Methods

Technical description of the Actiheart

The main component of the Actiheart is 7 mm thick with a diameter of 33 mm and houses a movement sensor, a rechargeable battery, a memory chip, and other electronics (Figure 1). A wire of approximately 100 mm length runs to a smaller ($5 \times 11 \times 22\,mm^3$) clip. The total weight is 8 g. The Actiheart is capable of measuring acceleration, HR, HR variability, and ECG amplitude for a set time resolution. Available epoch settings are 15 s, 30 s, or 1 min. The memory capacity of 128 kb allows the user to store data for more than 11 days with an epoch setting of 1 min. Another recording mode stores acceleration (15 s epoch) and all interbeat-intervals (IBI), that is, the time-intervals between 'R' spikes of the QRS complex for approximately 24 h, and finally the nonintegrated waveforms of acceleration (sampled at 32 Hz) and HR (sampled at 128 Hz) can be recorded for 13 min 38 s.

Movement measurement. Acceleration is measured by a piezoelectric element contained in the Actiheart with a frequency range of 1–7 Hz (3 dB). This movement sensor generates a transient charge, when exposed to time-varying acceleration and thus produces a voltage signal, which is then converted into a binary signal by an 8-bit A/D converter. This results in 256 distinctive levels of acceleration (128 positive and 128 negative levels). The accelerometer has a dynamic range of $\pm 25\,m/s^2$ (± 2.5 G) and its sensitivity per bit is $0.2\,m/s^2$ (0.02 G). The instantaneous

acceleration is quantified as numerical difference from zero acceleration in binary units, thus leaving a 2-bit ($\sim 0.4\,m/s^2$) wide deadband. The binary signal is stored in a cache 32 times a second and summed up over the epoch. At the end of the epoch, the sum is divided by 16 and then again by 2, *N* number of times until the number is below 32. The resulting integer and *N* are then stored in a single byte (5 bits for integer, 3 bits for *N*) in the nonvolatile memory and the cache is reset to zero. The movement sums are then divided by the calibration factor for the particular Actiheart unit during download by the software. Actiheart units are calibrated by the manufacturer with sinusoid accelerations of ± 1 G (average ~ 0.7 G), obtained with a calibration frequency of 3 Hz.

Heart rate measurement. The Actiheart has a sensitivity of 0.250 mV. The ECG signal is electronically amplified by a factor of 900 (amplifier frequency response: 10–35 Hz). The resulting ECG signal is sampled at 128 Hz and each R-wave decaying edge is identified by using the average difference of ECG samples n_i and n_{i+1} and ECG samples n_i and n_{i+2}. The threshold detection sensitivity changes with the amount of movement detected. At the end of the epoch, the trimmed mean of the last 16 R–R intervals is calculated by ignoring values outside $\pm 25\%$ of the initial mean. This is converted to beats-per-minute (bpm) and written to the memory at the end of each epoch. When the Actiheart is set up to record HR variability over the epoch, the two fastest and the two slowest interbeat intervals occurring in that epoch are also stored. The measurable range of HR in the manufacturer specification is 31–250 bpm.

Movement simulation study

This part of the study was designed to test the Actiheart accelerometer during sinusoidal movements in a controlled mechanical setting, which is described in detail elsewhere (Brage *et al*, 2003a). Intra- and interinstrumental reliability and validity were explored in eight Actiheart units, set up to record movement every 15 s. Data were obtained, using three different radius settings (*r*); 22.0, 35.5, and 49.0 mm through the frequency (*f*) range; 0.5–4.0 Hz at 0.25 Hz increments (0.5 Hz above 3 Hz). In this setup, the global average acceleration **A** is given by $\mathbf{A} = 8r\pi f^2$. The mechanical test series produced a total of 36 different accelerations, ranging from 0.1 to 19.7 m/s^2.

HR simulation study

Analyses of intra- and interinstrumental reliability of the HR sensor within the Actiheart were conducted with eight Actiheart units. The units were set up to record with an epoch of 15 s and then mounted in parallel on to strings of conductive tape to be tested with simulated R-waves (with amplitude 0.25 mV) on frequencies 25, 30, 33, 50, 100, 150, 200, and 250 bpm, for 2 min at each frequency.

Agreement with ECG during resting, walking, and running

The purpose of this part of the study was to assess the agreement between Actiheart, ECG, and Polar measurements of HR during both resting and treadmill exercise. Five men (32–46 y, 69–94 kg, 1.62–1.90 m, 22–30 kg/m^2) and four women (24–47 y, 59–64 kg, 1.58–1.71 m, 20–26 kg/m^2) were recruited. After light preparation of the skin, two ECG electrodes (Red Dot 2570, 3 M) were applied on the chest, with the medial electrode placed at the level of the third intercostal space on the sternum and the lateral electrode placed on the same horizontal level and as lateral as possible on the major pectoral muscle (Figure 1). The Actiheart was first set up to record waveforms for 2 min. If the HR trace was acceptable, that is, if the HR algorithm recognized the R-waves, we carried on with the protocol and set the Actiheart up to record in HR variability mode, storing activity every 15 s. Otherwise, we waited 5 min to allow skin impedance to drop and obtained another waveform measurement. The protocol was at least 4 min of resting followed by an exercise protocol of treadmill walking and running, consisting of 4 min of walking at each of the following speeds 3.2, 4.5, and 5.8 km/h and running at 8.5, 10.3, and 12.1 km/h or until volitional exhaustion. Three standard 3-lead ECG recordings (model 90369, Space Lab Medical) and three Polar HR readings (model S610, Polar Electro Oy, Kempele, Finland) were obtained during the last 45 s on each intensity level.

Prediction of PAI during walking and running

The purpose of this part of the study was to develop preliminary prediction equations for PAI during treadmill walking and running. Using the same protocol as described above, 11 men (29–50 y, 64–88 kg, 1.63–1.84 m, 21–29 kg/m^2) and nine women (26–45 y, 49–84 kg, 1.55–1.80 m, 20–29 kg/m^2) were recruited, in which we collected information on physiological energy expenditure. In this study, the Actiheart was set up to record in IBI logging mode, storing activity every 15 s. Oxygen consumption (VO_2) and carbon dioxide (VCO_2) production rates were measured breath-by-breath with a portable indirect calorimetry system (K4b^2, Cosmed s.r.l. Rome, Italy), previously validated (Hausswirth *et al*, 1997; McLaughlin *et al*, 2001). On all test days, gas analysers were calibrated against room air and a standard gas (16% O_2, 5% CO_2) before the first test, at least 45 min after the equipment had been turned on. Turbine, delay time, and room air calibrations were performed before each test. The weight of the K4b^2 equipment and the subject's footwear was set at 2.2 kg for all participants.

Before leaving the lab, the Actiheart was set up to measure movement, HR, and HR variability with an epoch of 1 min and the participants then wore it until the following morning to obtain a measure of sleeping HR.

A ramped step test, designed to account for some of the variance in individual HR–PAI relationships, was performed within 2 weeks of the treadmill test for each individual. The protocol involved 10 min of stepping up and down (=one cycle) a 215 mm high step, starting at 0.25 Hz (one foot placement per second) for the first minute and then steadily ramping up to 0.625 Hz at minute 10, followed by 1 min of recovery. The step test was administered in the form of a drum beat, included in the Actiheart software to facilitate time synchronization.

Informed written consent was obtained from all participants and the study was approved by the Cambridge Local Research Ethics Committee.

Statistics

All movement and HR data are expressed in counts per min (cpm) and beats per min (bpm), respectively. In both simulation setups, output values for each Actiheart unit were calculated as the mean of six epochs on each intensity setting. This yielded 36 means for each Actiheart unit in the movement simulation study and eight means per unit in the HR simulation study.

Intrainstrument reliability. For each Actiheart unit, the standard deviations of the calculated means were divided by their corresponding unit means to obtain coefficients of variation (CV_{intra}), used to assess intrainstrument reliability (low CV denotes good reliability). All values of CV_{intra} were plotted against average acceleration and HR for the movement simulation and HR simulation study, respectively, to examine how intrainstrument reliability varies in the investigated range.

Interinstrument reliability. The mean of the Actiheart unit means was also calculated for each acceleration setting (36 common means) and each HR setting (eight common means). Interinstrument reliability when testing multiple monitors is more complex but essentially the same as when assessing agreement between two (Bland & Altman, 1986). We explored this aspect in four ways: (a) by calculation of relative differences of the unit means to their common mean, which were plotted against acceleration and simulated HR, respectively, for each unit (modified Bland–Altman plots), (b) by calculation of the CV (CV_{inter}) of the common means, (c) with the Cronbach's alpha intraclass correlation coefficient (ICC), and (d) by two-tailed paired *T*-tests on the differences between unit means and common means to determine systematic bias (Bland & Altman, 1986; Atkinson & Nevill, 1998).

Technical validity of movement registration. Linear correlation analysis was performed on the relationship between average acceleration **A** and Actiheart movement counts. The strength of this association is an expression of the criterion-related validity.

Human observations. Actiheart movement counts and HR for the last three epochs (~45 s) of rest and on each treadmill level were averaged to yield steady-state values of movement and HR, respectively. Corresponding values were obtained from ECG and Polar HR data. Agreement between methods was assessed with limits of agreement, ICC, and paired *T*-tests (Bland & Altman, 1986; Atkinson & Nevill, 1998). With the exception of the HR agreement analyses, all HR values are expressed in bpm above sleeping HR (HRaS) (Andrews, 1971; Hiilloskorpi *et al*, 2003). Sleeping HR was defined as the 10th lowest observation during the night. The Actiheart movement flex point was calculated as the average between resting and slowest walking speed (Brage *et al*, 2004). The flex HR was calculated as the average between the sleeping HR and the HR during the slowest walking speed. The transition HR between walking and running was calculated as the mean HR between the fastest walking and the slowest running speed.

The breath-by-breath gas exchange data was filtered by disregarding ventilation outside the 3–80 breaths/min range, tidal volumes outside the 0.2–10 l/min range, expired O_2 fractions outside the 10–20% range, and expired CO_2 fractions outside the 1–10% range. After filtering, all data were smoothed with a running average (two neighbouring values). Average VO_2 and VCO_2 were calculated using the last minute of resting and each of the treadmill stages. Incomplete stages were not included in the analysis. Predicted maximal VO_2 was calculated by extrapolating the HR–VO_2 relationship to age-predicted maximal HR (MHR) (Tanaka *et al*, 2001). If age-predicted MHR or predicted maximal VO_2 values were surpassed during exercise, observed values were used. Predicted maximal VO_2 is expressed in ml/min. All other exercise values are expressed relative to body weight +2.2 kg, to account for weight of Cosmed equipment and shoes. Walking and running PAI was calculated as exercise intensity minus resting metabolic rate and expressed in J/min/kg (Weir, 1949). PAI was modelled from corresponding Actiheart data using multiple linear regression with clustering by individual to compute robust standard errors. Entered variables included age, sex, and the interaction between HR and sex. Exclusion of variables was carried out with the Wald statistic as the removal criterion ($P>0.10$). Comparisons between final models were performed with paired *T*-tests on (absolute) residuals.

Model refinements by step test. The mass-specific lift work rate (mechanical power) of the step test, P_{step}, was calculated as $P_{step}=9.81\,m/s^2 \times$ step height (m) $\times$ lift frequency (number of body weight lifts per min) and expressed in J/min/kg (note, that physical and physiological Joules are different). For each individual, linear regression was used to model P_{step} from HRaS, disregarding the first minute. This produced individual calibration parameters; HRaS slope and intercept (denoted β_{step} and α_{step}, respectively). We also did this disregarding the last 2 min of step data. The recovery HRaS during the first minute following the step test was

modelled as an exponentially decaying function according to the formula $HRaS(t) = HRaS_0\, e^{-t/\tau}$ where $HRaS_0$ is the HRaS at $t_{recovery} = 0$ min after termination of the step test and τ is the time constant of the step test HR recovery (halftime is thus given by $t_{1/2} = \tau \ln[2]$). The time constant τ characterises this decay and was estimated by regression on the logarithmically transformed HR trace for each individual, thus yielding a third individual calibration parameter. All three step test-derived calibration parameters (β_{step}, α_{step}, and τ) were then introduced into the two walking–running PAI equations containing the HRaS term.

In addition, the HR relationship with P_{step} was extrapolated to MHR to yield $Pmax_{step}$, which we used to derive a prediction equation for fitness, including also τ, age, sex, weight, height, and sleeping HR. All statistical analyses were performed with Stata 8.2 Intercooled version (Stata Corp, TX, USA).

Results

Movement simulations

The reliability of the accelerometer in the Actiheart was dependent upon acceleration magnitude. The median (range) CV_{intra} was 18 (0–245)% on accelerations below 0.7 m/s^2, whereas above this level, median (range) CV_{intra} was 0.0 (0–11)%. For the interinstrument reliability, median (range) CV_{inter} values were 89 (40–167)% below 0.5 m/s^2, 25 (17–33)% from 0.5 to 1.0 m/s^2, and 5.3 (4–14)% above 1 m/s^2, respectively. This heteroscedasticity was evident in the modified Bland–Altman plot of the relative unit differences against acceleration (Figure 2): Below 0.5 m/s^2, the geometric mean absolute and numerical percentage error of the eight Actiheart units was 11 cpm and 53%, respectively (unit differences range: 2–51 cpm; 3–358%), between 0.5 and 1.0 m/s^2 corresponding values were 12 cpm and 11% (range: 1–64 cpm; 1–70%), and above 1.0 m/s^2 values were 47 cpm and 4% (range: 0–728 cpm; 3–4%). Cronbach's alpha ICC was 0.9995 ($P<0.001$). Unit-specific systematic bias was statistically significant in all units ($P \leq 0.004$), ranging from −20 to 32 cpm which corresponds to −5 to 9% of the mean of all units in the whole acceleration range.

In terms of technical validity, the accelerometer output from the Actiheart was significantly related to average acceleration in a linear fashion, the relationship being $\mathbf{A} = 0.0029$ movement (cpm) + 0.37 ($R^2 = 0.999$, SEE = 0.16 m/s^2, $P<0.001$). Without intercept, this relationship was $\mathbf{A} = 0.0030$ movement (cpm) (SEE = 0.32 m/s^2).

HR simulations

The Actiheart did not pick up simulated HR at 25 bpm and returned 0 bpm but was otherwise within 1 bpm of the simulated HR values from 30 to 250 bpm. Median (range) values of CV_{intra} and CV_{inter} were 0.0 (0–3.3)% and 0.03 (0–0.9)%, respectively. Cronbach's alpha ICC of reliability was 0.993 ($P<0.001$).

Human observations: resting, walking, and running observations

Results of the HR agreement analysis are shown in Table 1. There were no significant mean differences between any of the methods. Numerical error between Actiheart and ECG was positively correlated with treadmill speed ($r = 0.30$, $P = 0.004$). The corresponding correlation between Polar and ECG was not significant ($P = 0.189$).

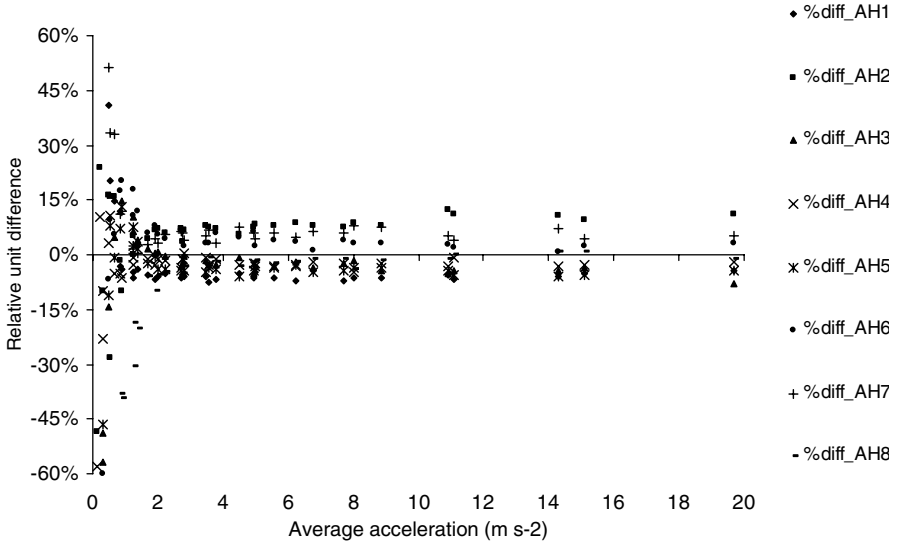

Figure 2 Interinstrument reliability of the movement sensor in the Actiheart: unit differences of eight Actiheart units expressed relative to their common mean and plotted against acceleration.

In the *energy expenditure study*, all participants ($n=20$) were able to complete the first four levels of exercise (all walking and the first running speed). All men and seven women also completed the second running speed, whereas nine men and five women completed the last running level. The men were on average 13.7 kg heavier ($P=0.006$) and 0.10 m taller ($P=0.005$) than the women. Age, BMI, sleeping HR, and resting VO_2 did not differ significantly between sexes. Mean (s.d.) resting VO_2 was 3.15 (1.22) ml/min/kg, which was not significantly different from 3.5 ml/min/kg (=1 MET).

Table 1 Agreement between ECG, Actiheart, and Polar S610 HR monitors ($n=9$)

	95% limits of agreement (bpm)	*Mean difference (bpm)*	*ICC*	*Heteroscedasticity (r)*
Actiheart—ECG	−4.2; 4.3	0.1	0.999	0.29*
Polar S610—ECG	−3.4; 4.3	0.4	0.999	0.08
Actiheart—Polar S610	−3.4; 2.6	−0.4	0.999	0.30*

ICC = intraclass correlation coefficient; heteroscedasticity = correlation between numerical error and mean HR.
*$P<0.05$.

Actiheart movement counts increased with treadmill speed (Figure 3) in a nonlinear manner; the slope being >10 times steeper in walking compared with running. This could be approximated with the linear equation Speed = 0.0029 · movement + 2.4 ($R^2=0.800$, $P<0.001$, SEE = 1.7 km/h). The mean (s.d.) movement flex point was 133.2 (49.1) cpm. HR increased linearly with speed (Figure 3), according to the equation Speed = 0.089HRaS + 0.11 ($R^2=0.887$, $P<0.001$, SEE = 1.3 km/h). Mean (s.d.) resting HRaS, flex HRaS, and walking/running transition HRaS were 10.2 (5.4), 23.6 (6.4), and 80.4 (14.3) bpm, respectively. Expressed as functions of sleeping HR, resting HRaS was −0.07Sleeping HR + 13.4, flex HRaS was 0.05Sleeping HR + 21.2, and transition HRaS was 0.54Sleeping HR + 54.3.

The relationship between PAI and treadmill speed (Figure 3) was PAI = 69speed–85 ($R^2=0.912$, $P<0.001$, SEE = 82 J/min/kg). Walking and running PAI models are displayed in Table 2. The combined model was significantly more accurate than each of the two single-measure models ($P<0.001$). Substituting HRaS with absolute HR was less accurate ($P\leq0.011$) in both relevant models.

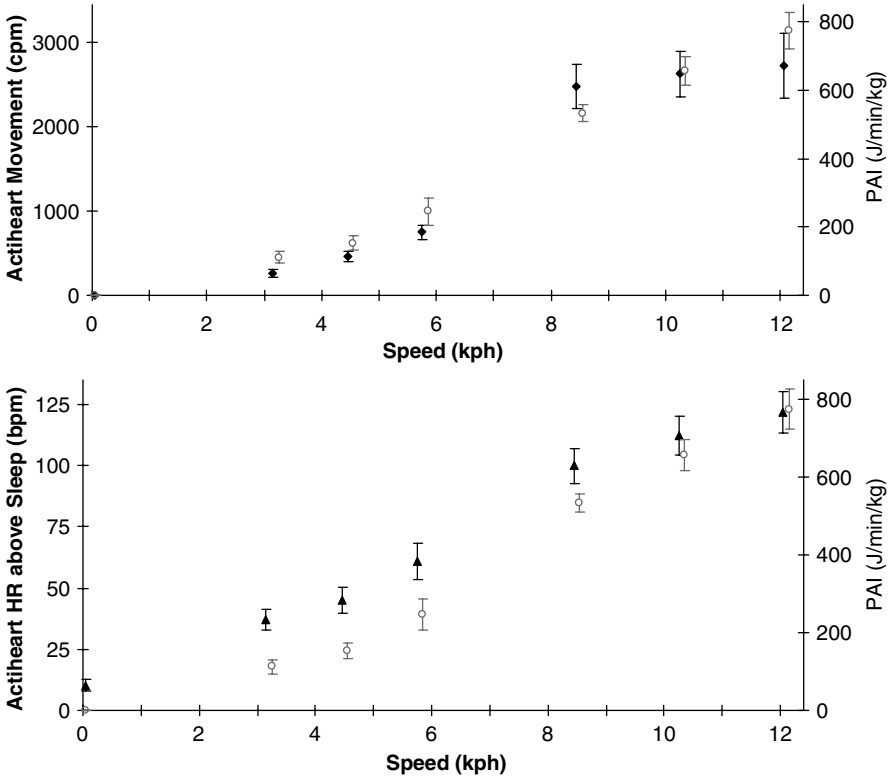

Figure 3 Data output from the Actiheart during human resting, walking, and running on left axes (black closed symbols) together with intensity on right axes (grey open circles). Upper panel is Actiheart movement in cpm and lower panel is Actiheart HR expressed in bpm above sleeping HR. Error bars are 95% confidence intervals.

Table 2 Prediction of walking running physical activity intensity from Actiheart

	PAI model (J/min/kg)	R^2	*SEE (J/min/kg)*
(a) Movement	0.21movement + 77male sex + 21	0.842***	109.9†‡§
(b) HRaS	5.5HRaS + 1.2HRaS male sex + 16male sex−94	0.903***	84.8†‡§
(c) Movement + HRaS	0.11movement + 2.3HRaS + 1.7HRaS male sex−17male sex−21	0.942***	65.7§
(d) HRaS w/step test	2.9HRaS + 1.1HRaS male sex + 2.9HRaS β_{step} + 1.3α_{step}−10male sex−75	0.937***	69.0§
(e) Movement + HRaS w/step test	0.081movement + 1.3HRaS + 1.5HRaS male sex + 2.0HRaS β_{step} + 1.2α_{step}−28male sex−22	0.956***	57.7†‡

HRaS = heart rate above sleeping HR; β_{step} = HR slope from step test; α_{step} = HR intercept from step test.
***$P<0.001$, †significantly ($P<0.05$) different from Model c, ‡significantly different from Model d, §significantly different from Model e.

Model refinements by step test

All participants were able to complete the step test. The relationship between HR and step power was $P_{step} = 0.41\text{HRaS} + 26.4$ ($R^2 = 0.45$, $P<0.001$, SEE = 10.7 J/min/kg) with individual Actiheart HRaS regression slopes (β_{step}) and intercepts (α_{step}) ranging from 0.58 to 1.3 J/kg/beat and from −37 to 13 J/min/kg, respectively. The average (range) recovery time constant τ was 2.1 (1.1–4.6) min. When step test parameters were introduced into the two PAI models containing HRaS (Table 2), model accuracies improved significantly ($P \leq 0.031$). Step parameters derived disregarding the last 2 min of stepping produced the most precise PAI models and are thus the models presented.

Predicted VO_2max (in ml/min) from the treadmill test could be estimated from step test parameters (8-min values), sex, weight, and sleeping HR, according to the model:

$$VO_2\text{max} = 9.4\text{Pmax}_{step} + 645\text{male sex} + 15\text{weight} - 23\text{Sleeping HR} + 1596$$

($R^2 = 0.896$, $P<0.001$, SEE = 303 ml/min).

The recovery time constant, τ, was rejected from this model ($P = 0.485$).

Discussion

Development of better methods for measurement of physical activity is a key research priority if we are to gain a comprehensive appreciation of the dose–response relationships of PA with mortality and morbidity (LaMonte & Ainsworth, 2001). In this paper, we have investigated various dimensions of the reliability and validity of a new combined movement and HR monitor, Actiheart. Measures of movement and HR generally agreed well with criterion measures of acceleration and HR, as well as provided relatively precise estimates of PAI during walking and running.

The technical evaluation revealed heteroscedasticity for the movement registration. Movement below 0.5 m/s^2 was measured with large relative error, whereas above 1 m/s^2 reliability was high. This may partially be due to the method used to compress and store the movement data; the maximum percentage error being when the initial sum is below 16, as this will be stored as '0' (100% error). This theoretical error declines to <8.6% for a stored count of 10, and <3.8% for a stored count of 25. Any observed error higher than these must be attributable to real differences between units. A similar error pattern has been reported for the MTI Actigraph using the same experimental setup and protocol, although interinstrument reliability of the MTI appeared to be somewhat poorer than that observed here for the Actiheart (Brage *et al*, 2003a). The significance of this error is likely to be smaller during human movement because of the rarity of *continuous* very low-intensity activity. In terms of technical validity, the Actiheart demonstrated a high linearity with acceleration, which is in contrast to the frequency-dependent validity of MTI Actigraph counts (Brage *et al*, 2003a). Such frequency response has been shown to inflate the between-individual variance of the relationship with walking and running intensity (Brage *et al*, 2003b, c).

Simulated HR was picked up by the Actiheart within 1 bpm above 30 bpm, which is comparable to newer HR monitors (Boudet & Chamoux, 2000). The agreement with ECG during rest and treadmill exercise ($n = 9$) was within 5 bpm for both the Actiheart and Polar S610, with errors being somewhat larger during higher intensity. This agrees well with observations made by other investigators (Karvonen *et al*, 1984; Leger & Thivierge, 1988; Seaward *et al*, 1990; Godsen *et al*, 1991; Terbizan *et al*, 2002). In our study, the Actiheart was placed on the participant's upper chest. HR readings may, however, be less prone to error if the Actiheart is positioned at the level just below the sternum where the Polar belt is normally placed and this issue should be addressed in future studies.

Positioning of the Actiheart also impacts on the movement registration. In this study, the Actiheart movement sensor measured along the longitudinal axis of the trunk. A tilt away from this axis would result in the attenuation of the registered movement signal if the acceleration along this new alternative axis is different from the acceleration along the longitudinal axis. For example, accelerations during walking and running are generally very low along the mediolateral axis of the body. Assuming that these accelerations are zero, a tilt of d degrees towards this axis would result in cosine[d] times lower movement counts, for example, 15 and 45° tilts would lower the counts by 3 and 30%, respectively. It is therefore important to ensure that the Actiheart is kept

reasonably horizontally oriented but without stretching the wire, as this could result in both discomfort and noisier HR readings during chest expansion.

The Actiheart movement response during walking and running was expected (Cavagna *et al*, 1976, 1988; Cavagna & Kaneko, 1977). In most individuals, walking posture of the trunk is vertical, whereas in running there is greater between-individual variance in the degree to which individuals lean forward when running commences and as running speed increases. Average vertical acceleration increase with walking speed but is approximately 1 G, regardless of running speed, whereas average horizontal acceleration (ie, the fluctuations in forward speed with each step) increase in a linear fashion over a wide range of running speeds (Cavagna *et al*, 1976, 1988; Cavagna & Kaneko, 1977). This is the reason why the slope of the speed–movement relationship is less steep within running values than within walking values. This levelling-off phenomenon has also been observed for other uniaxial accelerometers, thus highlighting the limited validity of vertical accelerometry as a measure of running intensity (Haymes & Byrnes, 1993; Brage *et al*, 2003c; King *et al*, 2004).

The PAI models were reasonably precise, with R^2 values that exceeded 0.84 and standard errors of the estimates between 85 and 110 J/min/kg (about 1.2–1.5 MET). Movement–PAI relationships have, however, been reported to be highly variable in diverse scenarios, such as during level and incline walking and running, stair climbing, cycling, stepping, and during load-bearing activities, (Montoye *et al*, 1983; Sekioka *et al*, 1998; Jakicic *et al*, 1999; Nichols *et al*, 1999; Bassett *et al*, 2000; Hendelman *et al*, 2000; Brage *et al*, 2003b, c; Ekelund *et al*, 2003; Strath *et al*, 2003; King *et al*, 2004; Treuth *et al*, 2004). Adding HR monitoring to movement registration may well decrease this source of error. Certainly, in the present study, the precision of the combined model of movement and HRaS exceeded that of the single-measure models ($R^2 > 0.94$, SEE < 66 J/min/kg or 0.9 MET) and also that reported for other activity monitors (Freedson *et al*, 1998; Jakicic *et al*, 1999; King *et al*, 2004). Precision of group-based HR equations for predicting PAI on the individual level is, however, challenged by between-individual differences, mainly in stroke volume, haemoglobin content, local blood flow regulation, and O_2 extraction (Li *et al*, 1993; Whipp *et al*, 1996; Strath *et al*, 2000; Rowell, 2004; Singel & Stamler, 2004). Some of these factors are different between sexes, which support our observation of a significant interaction between HR and sex. In addition, significant between-individual variance in the HR–PAI relationships is removed by expressing HR as beats above sleeping HR (Andrews, 1971; Hiilloskorpi *et al*, 2003). Nonetheless, residual error remains due to between-individual differences in response to exercise. As we demonstrate, it is possible to account for some of this residual error by including the step test parameters in the prediction equations for PAI; the SEE being significantly reduced by about 18%. Somewhat surprisingly, the most precise models included the 8-min step test parameters and rejected the recovery time constant τ. Although a shorter step test is more feasible, this observation may imply a need for tighter protocol adherence during the final stages of the test.

Alternative approaches to incorporate step test parameters into PAI models include modelling the slope and intercepts of the PAI relationships directly but this does not take into account the possible dependency between HR slope and intercept (Rennie *et al*, 2001). In either case, a different group of volunteers is needed to test the true validity of the derived equations (Bland–Altman scenario), as opposed to the criterion-related validity we report here.

Step test parameters also contributed significantly to the explained variance in VO_2max, and the derived model is comparable or slightly better than other indirect tests (Uth *et al*, 2004). However, due to correlated error between predictor and outcome variables, the ability of our model to estimate *true* VO_2max is likely less than is indicated by the high R^2 value. Nonetheless, the step test is easy to administer, even in a field setting, and may also be utilized to monitor individual changes over time. Since the Actiheart stores the nonintegrated data during the step test, the obtained exercise and recovery ECG may well have diagnostic and prognostic value (Cole *et al*, 1999; Mazur *et al*, 2003; Mora *et al*, 2003). Although utilization of individual sleeping HR and step test parameters to adjust the prediction equations derived in the present study broadens their generalizability to subjects with different characteristics, the equations are likely less valid for other activities than walking and running. A key future objective would therefore be to examine the validity of the Actiheart in more diverse scenarios. An issue of concern not resolved in the present study, is the necessary interpolation between resting and lowest walking intensity. By definition, PAI is zero at HRaS = 0 bpm and during periods of no movement. We suggest therefore that the two PAI equations are forced through the origin when movement and HRaS are below their respective flex points. Although validity of such equations for assessment of free-living PAI is unknown, we have previously demonstrated that precise estimates of PAEE (the time integral of PAI) in a simulated daily-living scenario can be obtained when treadmill-derived equations are combined in a branched model (Brage *et al*, 2004). The validity of this approach using the Actiheart, however, remains to be determined, preferably against free-living energy expenditure measured with the doubly labelled water technique.

In conclusion, Actiheart appears to be a reliable and valid tool for the measurement of movement and HR during standardized technical conditions and also in humans at rest and during walking, and running. As such, Actiheart has potential to more accurately assess physical activity energy expenditure than existing devices and may thus aid our understanding of the behavioural antecedents of metabolic dysfunction.

Acknowledgements
We like to thank the study participants for their time and effort. Mrs S Hennings, Mrs J Mitchell, Mrs S Emms, and Mrs E Lucia-Rolfe are acknowledged for assistance with data collection. The use of commercial names in this manuscript is for information purposes only and does not imply endorsement.

Conflict of interest: None to declare.

References

Ainslie P, Reilly T & Westerterp K (2003): Estimating human energy expenditure: a review of techniques with particular reference to doubly labelled water. *Sports Med.* **33**, 683–698.

Albanes D, Conway JM, Taylor PR, Moe PW & Judd J (1990): Validation and comparison of eight physical activity questionnaires. *Epidemiology* **1**, 65–71.

Alfonzo-Gonzalez G, Doucet E, Almeras N, Bouchard C & Tremblay A (2004): Estimation of daily energy needs with the FAO/WHO/UNU 1985 procedures in adults: comparison to whole-body indirect calorimetry measurements. *Eur. J. Clin. Nutr.* **58**, 1125–1131.

Andrews RB (1971): Net heart rate as a substitute for respiratory calorimetry. *Am. J. Clin. Nutr.* **24**, 1139–1147.

Atkinson G & Nevill AM (1998): Statistical methods for assessing measurement error (reliability) in variables relevant to sports medicine. *Sports Med.* **26**, 217–238.

Avons P, Garthwaite P, Davies HL, Murgatroyd PR & James WP (1988): Approaches to estimating physical activity in the community: calorimetric validation of actometers and heart rate monitoring. *Eur. J. Clin. Nutr.* **42**, 185–196.

Bassett Jr DR, Ainsworth BE, Swartz AM, Strath SJ, O'Brien WL & King GA (2000): Validity of four motion sensors in measuring moderate intensity physical activity. *Med. Sci. Sports Exerc.* **32**, S471–S480.

Bland JM & Altman DG (1986): Statistical methods for assessing agreement between two methods of clinical measurement. *Lancet* **1**, 307–310.

Boudet G & Chamoux A (2000): Heart rate monitors and abnormal heart rhythm detection. *Arch. Physiol. Biochem.* **108**, 371–379.

Brage S, Brage N, Franks PW, Ekelund U, Wong MY, Andersen LB Froberg K & Wareham N (2004): Branched equation modeling of simultaneous accelerometry and heart rate monitoring improves estimate of directly measured physical activity energy expenditure. *J. Appl. Physiol.* **96**, 343–351.

Brage S, Brage N, Wedderkopp N & Froberg K (2003a): Reliability and validity of the Computer Science and Applications accelerometer in a mechanical setting. *Meas. Phys. Edu. Exerc. Sci.* **7**, 101–119.

Brage S, Wedderkopp N, Andersen LB & Froberg K (2003b): Influence of step frequency on movement intensity predictions with the CSA accelerometer: a field validation study in children. *Ped. Exerc. Sci.* **15**, 277–287.

Brage S, Wedderkopp N, Franks PW, Andersen LB & Froberg K (2003c): Reexamination of validity and reliability of the CSA monitor in walking and running. *Med Sci. Sports Exerc.* **35**, 1447–1454.

Cavagna GA, Franzetti P, Heglund NC & Willems P (1988): The determinants of the step frequency in running, trotting and hopping in man and other vertebrates. *J. Physiol. (Lond)* **399**, 81–92.

Cavagna GA & Kaneko M (1977): Mechanical work and efficiency in level walking and running. *J. Physiol. (Lond)* **268**, 647–681.

Cavagna GA, Thys H & Zamboni A (1976): The sources of external work in level walking and running. *J. Physiol. (Lond)* **262**, 639–657.

Ceesay SM, Prentice AM, Day KC, Murgatroyd PR, Goldberg GR, Scott W & Spurr GB (1989): The use of heart rate monitoring in the estimation of energy expenditure: a validation study using indirect whole-body calorimetry. *Br. J. Nutr.* **61**, 175–186.

Cole CR, Blackstone EH, Pashkow FJ, Snader CE & Lauer MS (1999): Heart-rate recovery immediately after exercise as a predictor of mortality. *N. Engl. J. Med.* **341**, 1351–1357.

Ekelund U, Aman J & Westerterp K (2003): Is the ArteACC Index a valid indicator of free-living physical activity in adolescents? *Obes. Res.* **11**, 793–801.

Freedson PS, Melanson E & Sirard J (1998): Calibration of the Computer Science and Applications, Inc. accelerometer. *Med. Sci. Sports Exerc.* **30**, 777–781.

Godsen R, Carroll T & Stone S (1991): How well does Polar Vantage XL heart rate monitor estimate actual heart rate? *Med. Sci. Sports Exerc.* **23**, S14.

Haskell WL, Yee MC, Evans A & Irby PJ (1993): Simultaneous measurement of heart rate and body motion to quantitate physical activity. *Med. Sci. Sports Exerc.* **25**, 109–115.

Hausswirth C, Bigard AX & Le Chevalier JM (1997): The Cosmed K4 telemetry system as an accurate device for oxygen uptake measurements during exercise. *Int. J. Sports Med.* **18**, 449–453.

Haymes EM & Byrnes WC (1993): Walking and running energy expenditure estimated by Caltrac and indirect calorimetry. *Med. Sci. Sports Exerc.* **25**, 1365–1369.

Hendelman D, Miller K, Baggett C, Debold E & Freedson P (2000): Validity of accelerometry for the assessment of moderate intensity physical activity in the field. *Med. Sci. Sports Exerc.* **32**, S442–S449.

Hiilloskorpi HK, Pasanen ME, Fogelholm MG, Laukkanen RM & Manttari AT (2003): Use of heart rate to predict energy expenditure from low to high activity levels. *Int. J. Sports Med.* **24**, 332–336.

Jakicic JM, Winters C, Lagally K, Ho J, Robertson RJ & Wing RR (1999): The accuracy of the TriTrac-R3D accelerometer to estimate energy expenditure. *Med. Sci. Sports Exerc.* **31**, 747–754.

Karvonen J, Chwalbinska-Monet J & Säynäjäkangas S (1984): Comparison of heart rates measured by ECG and microcomputer. *Phys. Sports Med.* **12**, 65–69.

King GA, Torres N, Potter C, Brooks TJ & Coleman KJ (2004): Comparison of activity monitors to estimate energy cost of treadmill exercise. *Med. Sci. Sports Exerc.* **36**, 1244–1251.

Kriska A (2000): Physical activity and the prevention of type 2 diabetes mellitus: how much for how long? *Sports Med.* **29**, 147–151.

LaMonte MJ & Ainsworth BE (2001): Quantifying energy expenditure and physical activity in the context of dose response. *Med. Sci. Sports Exerc.* **33**, S370–S378.

Leger L & Thivierge M (1988): Heart rate monitors: validity, stability, and functionality. *Phys. Sports Med.* **16**, 143–151.

Li R, Deurenberg P & Hautvast JG (1993): A critical evaluation of heart rate monitoring to assess energy expenditure in individuals. *Am. J. Clin. Nutr.* **58**, 602–607.

Livingstone MB, Prentice AM, Coward WA, Ceesay SM, Strain JJ, McKenna PG, Nevin GB, Barker ME & Hickey RJ (1990): Simultaneous measurement of free-living energy expenditure by the doubly labeled water method and heart-rate monitoring. *Am. J. Clin. Nutr.* **52**, 59–65.

Luke A, Maki KC, Barkey N, Cooper R & McGee D (1997): Simultaneous monitoring of heart rate and motion to assess energy expenditure. *Med. Sci. Sports Exerc.* **29**, 144–148.

Masse LC, Heesch KC, Fulton JE & Watson KB (2002): Raters' objectivity in using the compendium of physical activities to code physical activity diaries. *Meas. Phys. Edu. Exerc. Sci.* **6**, 207–224.

Mazur W, Rivera JM, Khoury AF, Basu AG, Perez-Verdia A, Marks GF, Chang SM, Olmos L, Quinones MA & Zoghbi WA (2003): Prognostic value of exercise echocardiography: validation of a new risk index combining echocardiographic, treadmill, and exercise electrocardiographic parameters. *J. Am. Soc. Echocardiogr.* **16**, 318–325.

McLaughlin JE, King GA, Howley ET, Bassett Jr DR & Ainsworth BE (2001): Validation of the COSMED K4 b2 portable metabolic system. *Int. J. Sports Med.* **22**, 280–284.

Montoye HJ, Washburn R, Servais S, Ertl A, Webster JG & Nagle FJ (1983): Estimation of energy expenditure by a portable accelerometer. *Med. Sci. Sports Exerc.* **15**, 403–407.

Moon JK & Butte NF (1996): Combined heart rate and activity improve estimates of oxygen consumption and carbon dioxide production rates. *J. Appl. Physiol.* **81**, 1754–1761.

Mora S, Redberg RF, Cui Y, Whiteman MK, Flaws JA, Sharrett AR & Blumenthal RS (2003): Ability of exercise testing to predict cardiovascular and all-cause death in asymptomatic women: a 20-year follow-up of the lipid research clinics prevalence study. *JAMA* **290**, 1600–1607.

Nichols JF, Morgan CG, Sarkin JA, Sallis JF & Calfas KJ (1999): Validity, reliability, and calibration of the Tritrac accelerometer as a measure of physical activity. *Med. Sci. Sports Exerc.* **31**, 908–912.

Paffenbarger Jr RS, Brand RJ, Sholtz RI & Jung DL (1978): Energy expenditure, cigarette smoking, and blood pressure level as related to death from specific diseases. *Am. J. Epidemiol.* **108**, 12–18.

Rennie K, Rowsell T, Jebb SA, Holburn D & Wareham NJ (2000): A combined heart rate and movement sensor: proof of concept and preliminary testing study. *Eur. J. Clin. Nutr.* **54**, 409–414.

Rennie KL, Hennings SJ, Mitchell J & Wareham NJ (2001): Estimating energy expenditure by heart-rate monitoring without individual calibration. *Med. Sci. Sports Exerc.* **33**, 939–945.

Rennie KL & Wareham NJ (1998): The validation of physical activity instruments for measuring energy expenditure: problems and pitfalls. *Public Health Nutr.* **1**, 265–271.

Rowell LB (2004): Ideas about control of skeletal and cardiac muscle blood flow (1876–2003): cycles of revision and new vision. *J. Appl. Physiol.* **97**, 384–392.

Saltin B, Lindgarde F, Houston M, Horlin R, Nygaard E & Gad P (1979): Physical training and glucose tolerance in middle-aged men with chemical diabetes. *Diabetes* **28** (Suppl 1), 30–32.

Seaward BL, Sleamaker RH, McAuliffe T & Clapp III JF (1990): The precision and accuracy of a portable heart rate monitor. *Biomed. Instrum. Technol.* **24**, 37–41.

Sekioka K, Takaba H & Nakano T (1998): Parallel recording of physical activity on commercial Holter recorders. *Front. Med. Biol. Eng.* **8**, 253–268.

Singel DJ & Stamler JS (2004): Blood traffic control. *Nature* **430**, 297.

Speakman JR (1998): The history and theory of the doubly labeled water technique. *Am. J. Clin. Nutr.* **68**, 932S–938S.

Strath SJ, Bassett Jr DR & Swartz AM (2003): Comparison of MTI accelerometer cut-points for predicting time spent in physical activity. *Int. J. Sports Med.* **24**, 298–303.

Strath SJ, Bassett Jr DR, Thompson DL & Swartz AM (2002): Validity of the simultaneous heart rate-motion sensor technique for measuring energy expenditure. *Med. Sci. Sports Exerc.* **34**, 888–894.

Strath SJ, Bassett Jr DR, Swartz AM & Thompson DL (2001): Simultaneous heart rate-motion sensor technique to estimate energy expenditure. *Med. Sci. Sports Exerc.* **33**, 2118–2123.

Strath SJ, Swartz AM, Bassett DR, O'Brien WL, King GA & Ainsworth BE (2000): Evaluation of heart rate as a method for assessing moderate intensity physical activity. *Med. Sci. Sports Exerc.* **32**, S465–S470.

Tanaka H, Monahan KD & Seals DR (2001): Age-predicted maximal heart rate revisited. *J. Am. Coll. Cardiol.* **37**, 153–156.

Tate DF, Jackvony EH & Wing RR (2003): Effects of Internet behavioral counseling on weight loss in adults at risk for type 2 diabetes: a randomized trial. *JAMA* **289**, 1833–1836.

Terbizan DJ, Dolezal BA & Albano C (2002): Validity of seven commercially available heart rate monitors. *Meas. Phys. Edu. Exerc. Sci.* **6**, 243–247.

Treuth MS, Adolph AL & Butte NF (1998): Energy expenditure in children predicted from heart rate and activity calibrated against respiration calorimetry. *Am. J. Physiol.* **275**, E12–E18.

Treuth MS, Schmitz K, Catellier DJ, McMurray RG, Murray DM, Almeida MJ, Going S, Norman JE & Pate R (2004): Defining accelerometer thresholds for activity intensities in adolescent girls. *Med. Sci. Sports Exerc.* **36**, 1259–1266.

Uth N, Sorensen H, Overgaard K & Pedersen PK (2004): Estimation of VO_2max from the ratio between HRmax and HRrest—the Heart Rate Ratio Method. *Eur. J Appl. Physiol.* **91**, 111–115.

Wareham NJ, Jakes RW, Rennie KL, Mitchell J, Hennings S & Day NE (2002): Validity and repeatability of the EPIC-Norfolk Physical Activity Questionnaire. *Int. J. Epidemiol.* **31**, 168–174.

Wareham NJ & Rennie KL (1998): The assessment of physical activity in individuals and populations: why try to be more precise about how physical activity is assessed? *Int. J. Obes. Relat. Metab. Disord.* **22** (Suppl 2), S30–S38.

Weir JB (1949): New methods for calculating metabolic rate with special reference to protein metabolism. *J. Physiol.* **109**, 1–9.

Whipp BJ, Higgenbotham MB & Cobb FC (1996): Estimating exercise stroke volume from asymptotic oxygen pulse in humans. *J. Appl. Physiol.* **81**, 2674–2679.

Wickstrom G & Bendix T (2000): The 'Hawthorne effect'—what did the original Hawthorne studies actually show? *Scand. J. Work Environ. Health* **26**, 363–367.