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ORIGINAL ARTICLE

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Effect of combined movement and heart rate monitor placement on physical activity estimates during treadmill locomotion and free-living

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Abstract A placement effect on activity measures from movement sensors has been reported during treadmill and free-living activity. Positioning of electrodes may impact on movement artifact susceptibility as well as surface ECG waveform amplitudes and thus potentially on the precision by which heart rate (HR) is ascertained from such ECG traces. The purpose of this study was to examine the extent to which placement of the combined HR and movement sensor, Actiheart, influences measurement of HR and movement, and estimates of energy expenditure. A total of 24 participants (20–39 years, 45–109 kg, 1.54–2.05 m, 19–29 kg m^{-2}) were recruited. Whilst wearing two monitors, one placed at the level of the third intercostal space (upper position) and one just below the apex of the sternum (lower position), study participants performed level walking, incline walking, and level running on treadmill, and completed at least one day of free-living monitoring. Placement differences in HR data quality, movement counts, and energy expenditure (estimated from combined HR and movement) were analyzed with regression techniques. Quality of HR data was generally higher when monitors were placed in the lower position. This effect was more pronounced in men during both treadmill activity (relative risk, RR [95% CI] of noisy HR data in upper vs. lower position, RR = 1.3[0.3; 5.6] in women, RR = 174[14; 2,156] in men) and during free-living (RR = 1.2[0.4; 3.3] in women, RR = 25[9.6; 67] in men). There were minor placement differences ($\leq$ 8%) in movement counts only in women during incline walking and running. During free-living, no placement effect on counts was observed. In all test scenarios, estimates of energy expenditure from the two positions were not significantly different. Positioning the Actiheart at the level below the sternum may yield cleaner HR data. Regardless of which position is used, this has little or no effect on movement counts and energy expenditure estimates, which is encouraging for studies where research participants may have to position the monitors themselves.

Keywords Monitoring · Energy expenditure · Accelerometry · Body movement · Epidemiology · Measurement

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Introduction

Precise measurement of physical activity (PA) is important for characterizing the dose–response relationship between PA and health outcomes, for specifying which dimension of activity is most important, for comparisons between different cultural settings and time eras, and finally in lifestyle intervention studies to ascertain the extent to which the intervention was delivered or indeed accompanied by compensatory behavior in other activity domains (Goran and Poehlman 1992; Wareham and Rennie 1998). When measurement error of the exposure is unrelated to the outcome of interest (i.e., random error), the consequence is most likely an attenuation of the exposure–outcome dose–response relationship, whereas non-random error (bias) may give the observer the false impression that an association is either weaker or stronger than the true association. Since PA is a complex behavior, the scope for error in its measurement is great and results may potentially be misleading. Thus, it remains an important objective to pursue greater accuracy when assessing PA.

Combined heart rate (HR) and movement sensing has been suggested by many to offer greater measurement precision of PA than either method used in isolation (Avons et al. 1988; Brage et al. 2004, 2005;

Haskell et al. 1993; Moon and Butte 1996; Rennie et al. 2000; Strath et al. 2001 but the advantage of adding a measure of HR to movement registration would logically also depend on the precision by which HR per se is measured and vice versa. The cardiac activation sequence is a temporal and spatial function, an evolving wave of depolarization which propagates through the transfer matrix of tissues with varying conductive properties and geometry before it may be detected on the body surface. These matrix factors influence the ECG amplitude, which combined with noise elements affects the nature of the final observable signal between any two points on the surface (Horan et al. 1980; Li et al. 2003; Nash and Pullan 2005; Schijvenaars et al. 1995). For example, R wave amplitude from a 12-lead ECG is normally much higher in lead V5 than for example in lead aVL (Rautaharju et al. 1994), and in a study comparing nine electrode positions, covering the left chest area from the second rib to 20 cm below (sternum to the mid-clavicular line), highest amplitudes were most frequently found in the two lower quadrants (Burke et al. 2003).

Reliability and validity of the combined HR and movement sensor, the Actiheart, has been examined previously during treadmill walking and running on the level (Brage et al. 2005), where the monitor was placed at the level of the third intercostal space with the medial electrode close to the sternum and the lateral electrode somewhere on the major Pectoralis muscle, depending on the size of the individual. This muscle is involved in much of the upper body activity, individuals may engage in and this position may thus be relatively prone to movement artifact in the electric surface potential. Positioning the monitor at the level below the apex of the sternum (just below the level of V5) may be associated with less movement artifact along with higher ECG amplitudes, both of which may contribute to more reliable HR readings since commonly used mathematical algorithms may easily detect QRS complexes due to higher signal-to-noise ratio (Horan et al. 1980; Pan and Tompkins 1985; Schijvenaars et al. 1997; Searle and Kirkup 2000). Furthermore, positioning of the monitor below the sternum may be preferred by the wearer in some circumstances, e.g., because the instrument would be less visible. Since protocol compliance is likely to be enhanced if participant vanity is compromised to a lesser degree, this may be of special significance in population studies. It is unknown whether quantification of movement using the Actiheart is dependent on where the instrument is placed on the torso. Other investigators, however, have reported a placement effect for movement sensors placed differently on the hip (Welk et al. 2000) and lower back during treadmill activity but not during free-living (Yngve et al. 2003). The objective of this study was to determine the extent to which placement of the Actiheart influences HR data quality, movement counts, and physical activity intensity (PAI) and energy expenditure (PAEE) estimates during treadmill locomotion and free-living.

Methods

Participants

Twelve men (21–39 years, 66–109 kg, 1.70–2.05 m, 21–29 kg m^{-2}) and 12 women (20–34 years, 45–68 kg, 1.54–1.74 m, 19–26 kg m^{-2}) were recruited by posters and word of mouth from the Odense area, Denmark. All participants were Caucasian and reported to regularly engage in physical activity. They were asked to refrain from eating and vigorous intensity exercise for at least 2 h prior to arriving at the laboratory. All participants provided written informed consent. The study was approved by the local research ethics committee.

Study procedure

This study had a treadmill test component and a free-living component. For both of these components, four ECG electrodes (Red Dot 2570, 3 M), to which the Actiheart units are clipped, were applied to the participant's chest. One position was at the level of the third intercostal space (upper position) and the other was just below the apex of the sternum (lower position), as shown in Fig. 1. Before application, the electrode area was prepared by removing any body hair and gently rubbing the skin with a towel. Resting ECG amplitude in both positions were recorded.

Treadmill test

For all treadmill tests, the Actiheart units were set up to record the duration of every inter-beat interval (IBI) interval as well as movement counts every 15 s. All 24 study participants were tested at least once on the treadmill, and 18 participants were tested twice. The

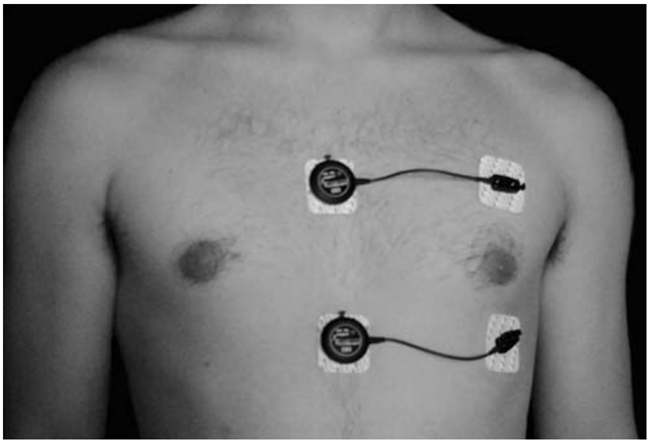

Fig. 1 Upper and lower positions for Actiheart attachment. The accelerometer is placed in the larger round clip orthogonally to the wire axis, thus here orientated to measure accelerations along the longitudinal axis of the body

treadmill protocol included a ramped stress test, which consisted of three main test phases. Phase 1 ('level walking') involved walking on the flat for 3 min at 3.2 km h^{-1}, then accelerating at 0.33 km h^{-1} per min for the next 6 min; phase 2 ('incline walking') consisted of brisk walking at 5.2–5.8 km h^{-1} for 6 min, with a progressively increasing gradient (at a rate of 1.7% min^{-1} for 6 min); and phase 3 ('level running') involved running on the flat with speed increasing from 9 to 13.8 km h^{-1} for 6.5 min (average acceleration 0.74 km h^{-1} per min) or until voluntary exhaustion. When participants were unable to complete the protocol, HR had reached at least 90% of age-predicted maximal HR (Tanaka et al. 2001).

Free-living observations

In this part of the study, the Actiheart units were set up to collect data in 'HR variability' mode with a measurement epoch of 15 s. This recording mode stores activity counts, average HR, ECG amplitude, the two longest and two shortest IBI intervals, and the number of ECG samples determined by the Actiheart firmware as 'LOST' (where a maximum of 128 samples/sec are possible) during calculation of the average HR per epoch. Participants wore two monitors for a median (range) duration of 26.0 (15.2–97.7) h. These data were also used for assessment of sleeping HR (SHR), and the calculation of HR above SHR (HRaS). Spare electrodes and basic replacement instructions were provided, just in case participants needed to detach monitors and electrodes.

Calculations and statistics

Sleeping heart rate was estimated as the HR below which 30 min of HR data was available in a 24-h period (disregarding observations where HR = 0 bpm). This method is relatively robust to estimation error but as a consequence may not necessarily represent the true minimum HR for each individual.

HR data quality for treadmill observations

We used a number of different definitions of 'HR noise' for the laboratory setting. For IBI logging data obtained during the treadmill test, the Actiheart reports '0 bpm' if the algorithm is unable to compute a HR value from the IBI intervals recorded during each 15 s epoch. All HR observations below SHR were regarded noisy, as were observations more than 175 bpm above SHR (denoted NOISE 1). This observation-by-observation evaluation of HR data quality represents a somewhat pragmatic approach. There are, however, more advanced options for capturing HR noise when employing a ramped exercise protocol because one may assume a relatively smooth increase in HR throughout the test. In the present study, we therefore also derived two regression-based noise definitions; the first (NOISE 2) were all observations more than 4 standard errors of the estimate (SEE) away from the expected HRaS to any given protocol time, derived from a third-order polynomial regression on the complete dataset (on observations regarded 'clean' by the NOISE 1 definition). Since it is expected that >99.9% of all observations would fall within this interval for any given time point in the protocol, simply by virtue of true between-individual variance, this definition would wrongly classify less than 0.1% of HR observations as noisy in the hypothetical situation that all were 'clean'. In the presence of low- or high-frequency noise, however, this noise detection algorithm would capture this, whereas observations which are truly noisy but happen to fall within this necessarily wide interval (because of true between-individual variance) would not be captured. Therefore, the second regression-based definition (NOISE 3) was performed on the individual basis, where true within-individual variance in the HR response to the employed protocol would be expected to be much less on an absolute scale. This approach, however, also limits the number of data points available for the regression, and so making it more difficult to obtain a good reference line. Therefore, the classification of a noisy HR recording by this approach was performed with a two-stage polynomial regression (third order) on single test(s) from each participant; the first stage being an initial flagging of potentially noisy HR data, defined as observations more than 2 SEE away from the expected HRaS to any given protocol time, derived by regression on each participant's data for each test and only using data not regarded noisy by the NOISE 2 definition. This first regression-based stage thus preliminarily captures (flags) observations that would otherwise greatly affect the regression line but most likely also filter out truly non-noisy recordings, since by definition about 5% of the observations would also be removed from a perfectly clean HR recording. The final stage is therefore a repetition of the regression from the first stage but this time only on the non-flagged data, thus producing a good reference for comparison. Based on this final regression, all observations were then re-evaluated and regarded noisy if they were not within 4.57 (trimmed) SEE of the individually estimated HR response. The assumptions underlying this double-pass filtering are, therefore, that once the overall HR response to the protocol is determined by a relatively robust two-stage regression, a decision is made on how much the observed HR can deviate from the predicted HR. Less than 0.1% of all HR observations would be expected to truly deviate by more than 4.57 trimmed SEE, corresponding to about 4 SEE on the original distribution so it is reasonable to assume these were noisy. Finally, we also subjectively reviewed each HR trace from the treadmill test and scored them for noise. This latter noise variable was only used to assess (apparent) sensitivity and specificity of

NOISE 1, 2, and 3. Summary results of noise are presented as noise rates (%), calculated as number of noisy observations divided by total number of observations. Logistic regression was used to compare noise rates between the two monitor positions and presented as relative risks (RR) with 95% confidence intervals (95% CI) using noise rate in the lower position as the reference. Noise classification (0 = clean, 1 = noisy) was entered as the dependent variable, and placement (0, lower positioned monitor; 1, upper positioned monitor) was entered as the independent variable.

HR data quality for free-living observations

We considered a couple of noise definitions when evaluating the free-living data. The first step, however, was to align the data for each pair of Actiheart monitors, so as to exclude periods when monitors were 'not worn'. Rather than using a 'monitor diary', we opted for an objective approach and defined such periods as no movement and HR = 0 bpm for more than 10 min occurring simultaneously in both monitors. Remaining data were then classified as noisy where HRaS was below −5 bpm and above 175 bpm (NOISE 1a). If for example SHR was estimated to be 50 bpm in one individual, then observations where HR < 45 bpm or HR > 225 bpm would be regarded as noisy by this definition. The reason for the margin of 5 bpm is that the estimate of SHR from a free-living HR recording does not necessarily return the absolute minimum HR, even if these were in fact real. Thus NOISE 1a may represent a somewhat conservative estimate of noise rate. In addition, we calculated noise in free-living by a second definition (NOISE 2a); by this definition, observations were regarded noisy when they differed more than 25 bpm from the moving average consisting of three consecutive (15-s epoch) values. The moving average was calculated disregarding HR observations which were noisy according to the NOISE 1a definition. For both of these binary definitions of noise (NOISE 1a and NOISE 2a), summary results are presented as noise rates (%), calculated and analyzed as described above for the treadmill data. For the free-living data, we also analyzed differences in HR data quality between the two positions using the 'LOST' variable (in %) as outcome. As this variable is continuous, multiple linear regression was employed for analysis. Placement (0, lower positioned monitor; 1, upper positioned monitor) was entered as the independent variable. Interaction with movement counts was also examined.

Effect of placement on movement counts, PAI, and PAEE estimates

For each treadmill phase and for free-living, effect of placement on both movement counts and PAI or PAEE estimated from combined HR and movement (Brage et al. 2005) using branched equation modeling was analyzed with multiple linear regression, adjusting for monitor. PAEE (expressed in kJ kg^{-1} day^{-1}) is the time-integral of PAI estimates (expressed in J kg^{-1} min^{-1}). The branched equation technique has been described in detail elsewhere (Brage et al. 2004). In brief, the technique is used to calculate a weighted average between a priori established movement-PAI and HR–PAI relationships. Based on immediate movement and HRaS, a decision tree assigns one of four different weightings to each epoch of data. For the present analyses, we used a branch-split movement value $X = 25$ cpm, and subsequent branch-split values $Y_1 = 0.54$, $Y_2 =$ 54.3 bpm, $Z_1 = 0.05$, and $Z_2 = 21.2$ bpm for HRaS. Weightings $P_{1\text{–}4}$ were set at 90, 50, 50, and 10%, respectively (Brage et al. 2005). Interactions with treadmill speed and phase were also examined. Epoch utilization of the branched model was calculated and differences analyzed with Chi-square test.

All analyses were conducted in STATA 8.2 (StataCorp, TX, USA) using clustering of standard errors on individual. Interaction between sex and placement was included and results are presented stratified by sex and combined.

Results

In total, 8,276 treadmill and 398,060 free-living observations in 24 subjects were analyzed. Resting ECG amplitude was 39% lower in women than in men ($P < 0.001$). A significant interaction between placement and sex was observed ($P = 0.001$) and stratifying by sex revealed resting ECG amplitudes to be 53% ($P = 0.014$) and 130% ($P < 0.001$) higher in the lower placement position in women and men, respectively. Mean (SD) SHR was 52.8 (7.3) bpm.

HR data quality

The average HRaS response to the employed treadmill protocol was HRaS(t) = $-0.013t^3 + 0.65t^2 - 4.2t + 39$ ($R^2 = 0.83$, SEE = 13.7 bpm) with protocol time measured in minutes. During treadmill activity, average noise rates were 4.2, 4.5, and 5.3% for NOISE 1, NOISE 2, and NOISE 3, respectively. The sensitivity and specificity of NOISE 1 was 80.7 and 100%, respectively, using the subjectively assessed noise as reference. Corresponding values for NOISE 2 and NOISE 3 were 86.6 and 97% (sensitivity), and 100 and 99.7% (specificity), respectively. We observed three times as much noise on the HR signal from monitors placed in the upper position, although this was only significant for men during all phases and only during the running phase for women and the combined analyses (Table 1). The placement effect on noise was more pronounced in men (RR = 174, [95% CI 14–2,156], $P < 0.001$ in men, RR = 1.3, [95% CI 0.3–5.6], $P = 0.71$ in women, $P < 0.001$ for interaction) during all phases. Further adjustment for electrode wear

Table 1 Noise rates in HR signal by placement and test scenario

	Women		Men		Combined	
	Upper	Lower	Upper	Lower	Upper	Lower
Level walking	0 (0–2.4)	0 (0–2.1)	0 (0–1.0)	0*** (0–0)	0 (0–1.4)	0† (0–0)
Incline walking	0 (0–2.8)	0 (0–7.4)	0 (0–0.7)	0*** (0–0)	0 (0–0.7)	0† (0–0.7)
Level running	24.0 (8.0–53)	0* (0–7.1)	9.7 (0–16)	0*** (0–0)	15.3 (2.1–42)	0***† (0–1.6)
Free-living	4.4 (0.9–6.5)	1.5 (0.1–6.2)	5.6 (1.9–11)	0.4*** (0.1–0.7)	5.3 (1.5–9.1)	0.5† (0.1–1.6)

Data are median noise percentages (inter-quartile range) of HR epochs regarded noisy (NOISE 3 for treadmill, NOISE 2a for free-living)
$^*P<0.05$, $^{**}P<0.01$, $^{***}P<0.001$, significantly different from noise rate at upper position as determined by multiple logistic regression
$^\dagger P<0.05$ for sex interaction

time did not alter these results. Resting ECG amplitude was inversely related to noise in all analyses ($P<0.05$) and adjustment for this variable in the placement analyses resulted in an attenuation of the placement coefficient. For women and only during the two walking phases, HR data from the upper position were significantly less noisy following adjustment for resting ECG amplitude. All results from analyses of HR noise during treadmill locomotion were similar regardless of which definition of noise was used.

During free-living, average noise rates were 5.36 and 5.44% for NOISE 1a and NOISE 2a, respectively. For free-living data, we observed a significant interaction between sex and placement both when using the binary noise variables ($P<0.001$, logistic regression) and the 'LOST' variable as noise indicator ($P=0.049$, multiple linear regression). There was no statistically significant effect of placement on HR quality in the women (RR of noise in upper vs. lower positioned monitor, RR = 1.2, [95% CI 0.4–3.3], $P=0.74$ for NOISE 2a and $\beta=0.013$, [95% CI −0.22 to 0.24], $P=0.90$ for LOST), whereas a significantly greater rate of noise was observed for the upper placement of the monitor in the men (RR = 25.4, [95% CI 9.6–67], $P<0.001$ for NOISE 2a and $\beta=0.46$, [95% CI 0.03–0.89], $P=0.037$ for LOST). Effect of placement was only borderline significant in combined analyses (RR = 2.4, [95% CI 0.9–6.4], $P=0.072$ for NOISE 2a and $\beta=0.018$, [95% CI −0.04 to 0.41], $P=0.11$ for LOST). Adjustment for movement, epoch ECG amplitude, or lab-measured resting ECG amplitude did not materially alter any of the above results. Epoch ECG amplitude was directly associated with noise in both men and women, whereas movement was directly associated with noise only in women. In contrast, lab-measured resting ECG amplitude was inversely associated with HR noise during free-living, although only significant in sex-combined analyses. Results were similar regardless of which definition of noise was used.

Effect of placement on movement counts

There was no placement effect on movement counts and no significant interaction between placement with either treadmill phase or speed (all phases). When each treadmill phase was analyzed separately, a significant placement effect was found for women during walking on the incline and running, with monitors placed in the upper position yielding 4% fewer counts and 8% higher counts than monitors placed in the lower position, respectively (Table 2). No placement effect on movement counts was found in men. In combined analyses, a significant placement effect (4% higher counts in upper position) and an interaction with sex was found during running. During free-living, there were no placement effects on movement counts in neither sex-stratified nor combined analyses (Table 2). This was also true when only analyzing non-zero count epochs.

Effect of placement on PAI and PAEE estimates

Table 3 shows the mean (± robust standard error, SE) PAI and PAEE by placement and test scenario estimated from the combined HR and movement data using branched equation modeling. Also displayed is the branch utilization of epochs. During treadmill locomotion (all phases analyzed together), average PAI was 8.6% higher when estimated from the lower positioned monitors when analyses were adjusted for monitor and sex ($P=0.012$). When each treadmill phase was analyzed separately, no significant placement effects were found. Similarly, there were no placement effects on average PAEE during free-living, and no significant interactions between sex and placement were observed in either scenario. Based on the observed variance, this study could have detected a difference of 10% in PAEE with 80% statistical power and a significance level of 5%. Proportions of epochs utilized by the four branches in the branched model used to estimate PAEE were also similar between placements.

Discussion

This study demonstrated a placement effect on HR data quality in men, whereas the findings in women were less clear. Furthermore, placement had an effect on movement counts during incline walking and running only in women but no placement effect was observed for the men during treadmill activity, nor during free-living for any of the sexes. Regardless of these effects on movement, there was no placement effect on PAI and PAEE estimated from combined HR and movement data.

Table 2 Movement counts by placement and test scenario

	Women		Men		Combined	
	Upper	Lower	Upper	Lower	Upper	Lower
Level walking	365 (9.5)	372 (8.1)	331 (9.6)	334 (8.3)	347 (6.9)	352 (5.3)
Incline walking	743 (21)	777* (13)	646 (27)	659 (18)	690 (17)	713 (10)
Level running	2,949 (76)	2,701** (65)	2,717 (88)	2,696 (74)	2,816 (57)	2,695*† (50)
Free-living	48.4 (4.0)	47.8 (4.1)	41.2 (5.0)	40.6 (5.0)	45.7 (3.3)	45.0 (3.4)

Movement count data are mean counts per min (robust SE)
* $P < 0.05$, ** $P < 0.01$ for placement difference as determined by multiple linear regression, adjusted for Actiheart unit
† $P < 0.05$ for sex interaction

Table 3 Physical activity intensity and energy expenditure by placement and test scenario

	Women		Men		Combined	
	Upper	Lower	Upper	Lower	Upper	Lower
Level walking	105 (7.8)	109 (8.9)	155 (5.1)	156 (5.4)	133 (4.2)	134 (5.4)
	2/98/0/0	5/95/0/0	4/96/0/0	0/100/0/0	3/97/0/0	2/98/0/0
Incline walking	246 (17)	264 (17)	261 (4.4)	266 (6.0)	254 (7.9)	265 (8.7)
	25/75/0/0	30/70/0/0	8/92/0/0	3/97/0/0	15/85/0/0	15/85/0/0
Level running	551 (20)	575 (19)	619 (12)	628 (13)	591 (11)	607 (10)
	93/7/0/0	94/6/0/0	94/6/0/0	92/8/0/0	94/6/0/0	93/7/0/0
Free-living	71.0 (5.3)	67.7 (4.5)	154.9 (3.0)	151.6 (3.9)	102.8 (3.7)	99.5 (3.2)
	5/23/21/51	4/25/20/51	4/19/22/55	1/21/17/61	4/21/22/53	3/24/19/54

Data are mean PAI (robust SE) in J min^{-1} kg^{-1} (treadmill) or PAEE in kJ day^{-1} kg^{-1} (free-living) and corresponding epoch utilization (percentage of total, branch 1/2/3/4) in the branched model (Brage et al. 2004)
* $P < 0.05$ for placement difference as determined by multiple linear regression, adjusted for Actiheart unit
† $P < 0.05$ for sex interaction

Our skin prep procedure reflected what we deemed acceptable to the general population. It is very likely that heavier skin preparation would result in less noise on the HR recordings (Fernandez and Pallas-Areny 2000; Klingler et al. 1979; Olson et al. 1979; Patterson 1978), but researchers would need to balance this against the acceptability of such procedures in the population, for which the method is intended to be applied. Moreover, for longer monitoring periods study participants may need to change electrodes and repeat the skin prep procedure. Related to this issue is which type of electrode to use, since conductive, adhesive, and allergenic properties are different between electrode types (Ask et al. 1979; Burke and Gleeson 2000; Fernandez and Pallas-Areny 2000; Geddes et al. 1962; Hagemann et al. 1985; Patterson 1978; Searle and Kirkup 2000; Xu et al. 1999). There is no published data on which type of electrode is the most suitable for the Actiheart but this may well be study specific.

An inverse association between HR data noise and resting ECG amplitude obtained by waveform in the lab was observed but a direct association between noise and the epoch-by-epoch stored ECG amplitude was observed during free-living. The latter was somewhat unexpected as R-wave detection is more difficult when ECG amplitude is low, and therefore by implication one would expect greater error in the measure of HR stemming from the trimmed average of the inter-beat intervals in each epoch. The reason for this discrepancy may be that sources of noise, e.g., movement artifacts, may affect not only the estimate of HR but also the measured ECG amplitude.

Noise rates were generally higher in women, which would be explained by either lower ECG amplitudes at heart level, greater electrical damping (attenuation) of the signal when traveling through the transfer matrix to the surface of the skin and electrode, and/or higher degree of sources of noise, e.g., movement artifacts. Signal attenuation due to different electro-conductive properties of breast tissue is unlikely, since we observed this gender difference in both positions. Moreover, breast protuberance has been shown to have only small effects on ECG amplitudes (Rautaharju et al. 1998). It is more likely that the greater attenuation is explained by higher level of subcutaneous fat, since adiposity has been shown to impact on surface ECG amplitudes (Rautaharju et al. 1994). In addition, movement artifacts, especially during more vigorous activities such as running, would expectedly be higher in women in both monitor positions, possibly depending on type of clothes being worn, e.g., sports bra. The less pronounced placement effect on HR noise in women compared to men could be explained by a differentially different effect of movement artifact in the two positions and/or less attenuation *difference* of the signal (153% in lower relative to upper position) in women compared to the attenuation *difference* (230% in lower relative to upper position) in men. It is also worth noting that the noise level in the lower position in the men was extremely low, which explains the high relative risk estimates, e.g., RR = 174, in men.

There was a tendency for noise to be more pronounced when movement counts were higher, although only significant in the women. This is somewhat concerning, since most analytical models of combined HR and movement depend more on using the HR data during higher intensities of physical activity, that is when there is most likely also a lot of movement (Brage et al. 2004; Rennie et al. 2000; Strath et al. 2001, 2002). However, estimates of PAI and PAEE using published prediction equations (Brage et al. 2005) in combination with a branched equation modeling approach (Brage et al. 2004) were not significantly different between the two positions during both treadmill locomotion and free-living, indicating that placement of the monitor has little or no impact on the estimated PAEE from combined sensing. This was despite the fact that some placement effects on movement counts were observed during incline walking and running in women (in opposite directions) and during running in the combined analyses. Apart from these three analyses, there was no evidence that positioning of the monitor made any difference to the movement output. One possible explanation for registering higher movement counts during running might be a more pronounced curvature of the spine, which would increase the relative contribution from the horizontal (forward–backward) acceleration component. This component increases with increasing running intensity, whilst the vertical component remains relatively constant (Cavagna et al. 1976; Chang and Kram 1999). To our knowledge, there are no other studies which have assessed the effect of placement on accelerometers attached to the two positions used in this study but Welk and colleagues reported a placement effect for the uni-axial CSA accelerometer (also known as the MTI Actigraph) but not for the tri-axial Tritrac and the tilted (45-degree from vertical) uni-axial Biotrainer accelerometers when placed differently on the hip during treadmill walking (Welk et al. 2000). Yngve and co-workers reported placement differences in CSA output placed on the hip and lower back during treadmill and over-ground walking and running, but this difference was non-significant during free-living (Yngve et al. 2003).

One limitation of the present study was that golden standard ECG was not available to perform agreement analysis with Actiheart measured HR in the two positions, although the HR algorithm has been successfully validated against ECG in a previous study (Brage et al. 2005). This would have allowed comparison of true sensitivity and specificity between positions. Likewise, the specific definitions of noise used in present study may not be universally applicable but some principles may well generalize to other study scenarios. Nonetheless, these limitations will not have any impact on the comparison in HR data quality between positions, as the same algorithms were applied to all data. The same is true for the comparison of PAEE estimates obtained through branched equation modeling. Another limitation is that we cannot be absolutely sure that our study participants kept both monitors in the allocated positions during the free-living part of the study. Furthermore, our results may not be generalizable to people who are anatomically much different from the current sample, e.g., obese individuals.

In summary, positioning the combined HR and movement sensor at the level below the sternum is associated with cleaner HR data, particularly in men. There may be a small placement effect on movement counts during more vigorous treadmill activity but this has little or no consequence for the physical activity intensity and energy expenditure estimated from combined HR and movement during treadmill and free-living activity. The latter is of special interest when combined monitors are used in studies where research participants may be expected to position the monitors themselves.

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