Gravitational artefact in frequency spectra of movement acceleration: implications for actigraphy in young and elderly subjects

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Abstract

Actigraphy, the long-term assessment of wrist movements by means of a small solid-state recorder, is widely used in a variety of human research fields, among which sleep, circadian rhythms and aging. Actigraphs assess movement with the use of accelerometers, which sense accelerations resulting from muscle force as well as accelerations due to changes in the position of the sensor in the gravitational field. In the present paper a method is described to minimise gravitational artefact in movement assessment by calculating the instantaneous acceleration vector from 3 perpendicular acceleration signals. It is shown that the power spectra of single axis acceleration signals are dominated by low-frequency components (± 0.25 Hz) due to gravitational artefact. Spectra of the instantaneous acceleration vector indicate that 'true' movement accelerations resulting from muscle force are present in a much wider range: from 0.25 to 11 Hz. Wrist accelerations in elderly subjects were found to be of lower amplitude and frequency as compared to young subjects. It is furthermore shown that a bandpass filter of 0.25 to 2 or 3 Hz, as has been used in commercially available actigraphs, is far from optimal, and may even result in a positive bias for movement detection in the elderly. This bias may underly contradictory findings in actigraphic studies on human aging. When a bandpass filter of 0.5–11 Hz is applied to a single-axis acceleration signal, the influence of gravitational artefact and bias are minimized, and the age-related decline in activity is properly detected.

Keywords: Aging; Activity; Actigraphy; Accelerometry; Bandpass filtering; Movement analysis; Gravitation; Sleep; Circadian rhythm

1. Introduction

Actigraphy, the long-term measurement of movements with a small wrist-worn device, is widely used in a variety of human research fields, among which sleep, circadian rhythms, aging, depression, hyperactivity and obesity (Tryon, 1991). Actigraphy is easy to apply and is tolerated even by very uncooperative subjects like demented elderly people (Van Someren et al., 1995). In commercially available actigraphs movement accelerations are sensed by an accelerometer, the output of which is amplified, bandpass filtered and fed to a threshold detector. Subsequently, 'movement counts' are obtained by increasing a counter either with every threshold crossing or with every sample that is above threshold. After the elapse of a programmable interval of time (e.g., 1 min or 1 h), the summed move-

ment counts are stored in memory, the counter is reset to zero, and counting is started for the next time interval. After completion of the registration, which can continue for weeks, the memory, containing movement counts for every time interval, is read out by a computer and the resulting time series can be plotted and analysed. Several variables can be derived, e.g., circadian phase and amplitude and day- and nighttime activity level. These variables have been shown to be useful in the description of activity during sleep and wakefulness in several disorders. In actigraphic studies on normal human aging, however, remarkably contradicting results have been reported. Whereas an age-related decrease in activity level is found in animal studies (cf., Aschoff, 1994), human aging studies using actigraphy have reported a prominent decrease (Van Hilten et al., 1995), a small decrease (Renfrew et al., 1987), no change (Witting et al., 1990), and even an increase (Lieberman et al., 1989) in activity level with aging. Aschoff suggested measuring speed of movements in addi-

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tion to movement counts in order to clarify the contradictory results (Aschoff, 1994).

At present, a new type of actigraph is being developed (Van Someren et al., 1993), primarily for use in studies of aging and neurodegenerative diseases. In addition to the measurement of the time above threshold of movement accelerations, this actigraph assesses average acceleration and discriminates tremor from other movements. In the course of this development, we re-evaluated how filters conditioning the acceleration signal should be chosen for optimal movement detection in young and elderly subjects. Commercially available actigraphs have applied a band-pass filter of 0.25 to 2 or 3 Hz, a choice that appears to be based on the work of Redmond and Hegge (1985). They concluded that accelerations caused by wrist movements are predominantly of very low frequency (0.25 Hz). This conclusion can be questioned for several reasons. First, the 0.25 Hz predominance implies that a characteristic movement will take 4 s, which is quite counter-intuitive. Second, studies on writing movements report frequency components up to 10 Hz (Teulings and Maarse, 1984). Third, in an attempt to discriminate normal movements from tremor in patients with Parkinson's disease by means of filtering (< 3 Hz normal; > 3 Hz tremor), we found accelerations of healthy subjects to contain so many signals greater than 3 Hz that an alternative method for discrimination had to be developed (Van Someren et al., 1993).

Re-evaluating the work of Redmond and Hegge (1985), we suspected the 0.25 Hz frequency peak to represent the acceleration signal caused by positional changes of the sensor in the gravitational field rather than by movement force. Repositioning of the axes of the sensor in the gravitational field may induce a strong signal in an accelerometer: a 180° rotation may even result in a signal change of 2 g, which appears to be high in comparison with the acceleration expected from movement force. Less extreme repositioning of the axes may still strongly affect power spectra derived from the signal: when the orientation of the sensor at the beginning of the time interval differs from the orientation at the end of the time interval, the resulting trend induces a strong peak at the first frequency component of the power spectrum, that may completely obscure other frequency components due to power leakage (Stone et al., 1992).

In a single-axis acceleration signal the contribution of gravitation and muscle force can not be separated. The contribution of gravitation can be reduced, however, by assessing movements using an accelerometer with 3 perpendicular axes, and calculating the instantaneous acceleration vector (Frost, 1978):

$$A(t) = \sqrt{x(t)^2 + y(t)^2 + z(t)^2},$$

where $A(t)$ is the instantaneous acceleration vector at time t , and $x(t)$, $y(t)$ and $z(t)$ are the single-axis accelerations in 3 perpendicular planes at time t . Whereas a change in the

Fig. 1. Results of a computer simulation study of a movement resulting in a sinusoidal signal in the x and y axes. Superimposed is a slow orientation change of the sensor in the gravitational field resulting in a shift of the 1 g contribution from the y axis to the x axis. For reasons of simplicity, no movement or gravitational forces in the z direction were implemented in this example. From top to bottom, the x , y , and A time series are shown in the left panels, and their power spectrum in the right panels. Note that the influence of the orientation shift on the instantaneous acceleration vector is limited to a very slight amplitude modulation, and that the 5 Hz peak representing movement acceleration is overshadowed by a low-frequency component in the single-axis spectra, but not in the instantaneous acceleration vector spectrum.

orientation of the axes of the sensor in the gravitational field strongly affects the single-axis signals, the contribution of gravitation to the spectra of the instantaneous acceleration vector is very limited. An example of this is given in Fig. 1, showing the results of a computer simulation study. In this simulation a sinusoidal movement signal with an amplitude of 0.3 g in the x direction and 0.1 g in the y direction was generated. Superimposed on these signals was a slow shift in the gravitational contribution of 1 g; from complete presence in the y direction to complete presence in the x direction, simulating a 90° positional change of the sensor. As is clear from Fig. 1, the spectra of x and y are dominated by the slow (gravitational) component, whereas the instantaneous acceleration vector does not show this component.

In the present study wrist movement accelerations in young adults and elderly subjects performing a short series of tasks were assessed in order to investigate: (1) whether the low-frequency component of single-axis movement acceleration signals is also present in the instantaneous acceleration vector, or rather an artefact of changes in the position of the sensor in the gravitational field; (2) whether the frequency distribution of the instantaneous acceleration vector differs in young and elderly subjects; and (3) how the choice of a bandpass filter in single-axis recordings affects age-related changes in wrist activity.

2. Methods

Fifteen healthy young adults (7 female, 8 male; age: mean $\pm$ SD 28 ± 4) and 10 healthy elderly subjects (6 female, 4 male; age: 80 ± 9) wore a small triaxial accelerometer (Entran EGS3-A-2, size $25 \times 25 \times 21$ mm, weight 50 g) on their dominant wrist while performing a series of tasks. The wrist was chosen as it is used most frequently as the place of attachment of actigraphs. Attachment to the waist, as has also been applied in some studies, may contain less gravitational artifact because positional changes in the gravitational field are thought to be less frequent as compared to wrist attachment. The axis transverse on the wrist was defined as x , the axis from the hand to the arm as y and the axis perpendicular to the wrist as z . The accelerometer signals were recorded on 3 channels of a portable analog recorder (TEAC HR-30 J) worn at the waist. The recording system had a linear sensitivity from DC to 39 Hz. Six tasks were performed: (1) 50 s (5×10) of finger tapping, (2) 30 s of diadochokinesis, (3) ± 45 s of placing pins in holes, (4) 30 s of tooth-brushing, (5) ± 3 min of writing and (6) a 5 min walk. Start and stop times of the recording and tasks were timed precisely. The first 2 tasks represent extreme movements that are normally assessed only in a clinical or experimental setting, and the latter 4 tasks represent more natural movements.

A flowchart of the steps in data analysis is shown in Fig. 2. The recordings were played back on a TEAC R71 at 32 times the recording speed, and sampled with a resolution of 12 bits (4096 levels with ± 2048 corresponding to ± 2 g at the sensor) and a sampling speed of 4096 Hz, corresponding to a sampling frequency of 128 Hz for the original recordings. Data analysis was subsequently performed on a PC. To correct for offset and gain differences in the 3 channels, the system was calibrated using recordings of the 6 planes of the sensor placed flat on a vibration-free horizontal table. After correction, offset was zero and 1000 AD units corresponded to 1 g. A 2-step, lag 1 moving average (Ruchkin and Glaser, 1976) was applied to the signal for noise reduction. From the recorded perpendicular x , y and z signals the instantaneous acceleration vector (A) was calculated as the square root of the sum of the squared x , y and z sample values (Frost, 1978). Subsequently, the Fast Fourier Transform was used to calculate the power spectral density and normalized power spectral density on 4-s epochs of x , y , z and A . Using the precisely recorded timed individual start and stop times of the tasks, the spectral densities were averaged over epochs belonging to the same task.

First, in order to investigate the presence of low-frequency components in single-axis and instantaneous acceleration signals, the normalized power spectral density of x , y , z and A were averaged for young and elderly subjects and plotted for each of the 6 tasks. Spectra were normalized by dividing each component by the sum of the spectral components. By normalizing, the frequency distri-

Fig. 2. Flowchart of the steps in data analysis.

bution of every subject contributed equally to the average, no matter whether the individual's movements were intense or small.

Second, in order to investigate age differences in wrist movements, 3 variables were derived from the spectra of the instantaneous acceleration vector for each subject and task: peak frequency (1), peak power (2) and (3) total power in the 0.25 to 11 Hz range. (1) and (2) were determined after applying a single lag 1 moving average (Ruchkin and Glaser, 1976) smoothing. In order to assess actual power, reflecting movement intensity, the non-normalized spectra were used for these calculations. For each task, differences between young and elderly subjects on these variables were evaluated using Mann-Whitney tests

with a 2-tailed significance level of $\alpha = 0.05$. A non-parametric test was chosen because of deviation from normality and differences in variance.

Third, as the instantaneous acceleration vector is not easily implemented in a low-cost actigraph, it was investigated how the choice of the bandpass filter for a uni-axial

Fig. 3. Average normalized power spectral density plots for young adults (dashed lines) and elderly subjects (solid lines). Horizontal axes show the frequency from 0 to 11 Hz, vertical axes the normalized power spectral density. Columns indicate the acceleration planes, from left to right x , y , z and A . Rows indicate the task during which the spectra were derived, from top to bottom: FTT = finger tapping; DDK = diadochokinesis; PIN = placing pins in holes; BRU = tooth-brushing; TXT = writing and WALK = walking. Note that most single-axis spectra (x , y and z) are dominated by a low-frequency component that represents gravitational artefact, and is absent in the spectra of the instantaneous acceleration vector (A , right column).

Fig. 4. Peak frequency (left column, in Hz), peak power (middle column, in g^2) and total power (right column, in g^2) derived from the power spectral density of the instantaneous acceleration vector during 6 tasks (rows; see legend of Fig. 3). Individual values are indicated by dots, and the group means of young adults and elderly subjects are indicated with a horizontal line. Note that peak frequencies occur throughout the range of 0.25–11 Hz.

signal affects the resulting power in the x , y and z signals. For every subject and task the (non-normalized) x , y and z power spectral densities in 2 frequency bands used in currently available actigraphs and in 1 alternative frequency band were calculated: (1) bandwidth 0.25–2.00 Hz, as applied in the actigraphs of Ambulatory Monitoring Inc.; (2) bandwidth 0.25–3.00 Hz, as applied in the actigraphs of Hombrechtikon; and (3) bandwidth 0.5–11 Hz, as applied in an actigraph that is presently being developed by our group (Van Someren et al., 1993).

Differences between young and elderly subjects in the summed spectral power in these ranges were evaluated using the Mann-Whitney test with a 2-tailed significance level of $\alpha = 0.05$.

3. Results

Activity was recorded in all subjects on almost every task. Assessment of walking was missing in 2 elderly subjects who felt uncomfortable about walking around with the equipment. The diadochokinesis recording of one elderly subject was discarded as the subject did not properly follow the instructions.

Fig. 3 shows the average normalized power spectra of the 6 movement types in young and elderly subjects in the x , y , z and A dimensions. The plots show clearly that low-frequency components dominate most of the single-axis signals, while their presence in the instantaneous acceleration vector A is limited, indicating that the strong presence of these components in single-axis spectra is due to changes of the sensor in the gravitational field.

Fig. 4 shows the peak frequency (left), peak power (middle) and total power (right) in the spectra of the instantaneous acceleration vector for each task. Individual values (dots) indicate a substantial between-subject variability throughout a frequency range of 0.25–11 Hz. Group means are indicated by horizontal lines, and Table 1 shows that these means differ significantly between young and elderly subjects on most tasks, indicating that movement accelerations of young subjects are of a higher frequency

and amplitude than the movement accelerations of elderly subjects. During finger tapping, however, a non-significantly higher power was found in elderly subjects as compared to young adults. This may be the result of a difference in the way the task was performed. Although not systematically assessed, we had the impression that more elderly subjects tapped with the use of their arm, whereas more young adults tapped with the use of a finger only.

Table 2 shows how the choice of the bandpass filter is of influence in discriminating single-axis movement accelerations in young and elderly subjects. The left columns give the average power in the 0.25–2 Hz range, in the x , y and z direction. This range has been applied in the actigraphs of Ambulatory Monitoring Inc. Depending on the plane of sensitivity of the sensor, which has not been documented for this actigraph, significantly lower wrist activity power in elderly people is only found during walking, and even significantly higher power can result from finger tapping and writing. The middle columns give the average power in the 0.25–3 Hz range, as has been applied in the Hombrechtikon actigraph. Depending on the plane of sensitivity of the sensor, which has also not been documented for this actigraph, significantly lower wrist activity power in the elderly is again limited to movements during walking, and even significantly higher power can result during finger tapping, tooth-brushing and writing. The right columns give the average power in the 0.5–11 Hz range, as applied in an actigraph that is presently being developed by our group (Van Someren et al., 1993). In a prototype of this actigraph the sensor is sensitive to accelerations in the z plane. In this direction lower wrist activity power in elderly subjects is found during diadochokinesis, pin placement, tooth-brushing and writing. In none of the tasks is wrist activity power in elderly subjects significantly higher than in young subjects. Applying a 0.5–11 Hz bandpass filter on a single (z) axis wrist movement signal is thus in better agreement with the finding of age effects in the instantaneous acceleration vector than the bandpass filters that have been applied in commercially available actigraphs.

Table 1

Group means, standard deviations and significance (Mann-Whitney) of the difference between young and elderly subjects for the peak frequency (left columns, in Hz), peak power (middle columns, in g^2) and total power (right columns, in g^2) derived from the power spectral density of the instantaneous acceleration vector during 6 tasks (rows; see legend of Fig. 3)

Task	Average ($\pm$ SD) peak frequency			Average ($\pm$ SD) peak power			Average ($\pm$ SD) total power		
	Young	Aged	P	Young	Aged	P	Young	Aged	P
FTT	7.0 $\pm$ 2.1	3.4 $\pm$ 2.7	< 0.002	0.003 $\pm$ 0.004	0.004 $\pm$ 0.003		0.017 $\pm$ 0.024	0.027 $\pm$ 0.028	
DDK	4.9 $\pm$ 1.8	4.8 $\pm$ 2.3		0.038 $\pm$ 0.016	0.019 $\pm$ 0.014	< 0.007	0.385 $\pm$ 0.129	0.270 $\pm$ 0.159	< 0.03
PIN	2.7 $\pm$ 0.6	2.1 $\pm$ 1.3	< 0.04	0.003 $\pm$ 0.002	0.002 $\pm$ 0.002	< 0.10	0.044 $\pm$ 0.012	0.32 $\pm$ 0.017	< 0.06
BRU	4.5 $\pm$ 0.5	3.8 $\pm$ 0.6	< 0.004	0.025 $\pm$ 0.022	0.012 $\pm$ 0.013	< 0.06	0.143 $\pm$ 0.107	0.076 $\pm$ 0.052	
TXT	4.4 $\pm$ 2.6	1.6 $\pm$ 2.3	< 0.01	0.0002 $\pm$ 0.0001	0.0003 $\pm$ 0.0003		0.004 $\pm$ 0.002	0.004 $\pm$ 0.002	
WALK	1.7 $\pm$ 0.4	1.5 $\pm$ 0.2	< 0.06	0.013 $\pm$ 0.004	0.008 $\pm$ 0.007	< 0.03	0.087 $\pm$ 0.024	0.050 $\pm$ 0.025	< 0.01

Only significance levels up to 0.10 are shown. Note that all significant differences and trends indicate a lower frequency and power in elderly subjects (marked with <).

Table 2
Group means, standard deviations and significance (Mann-Whitney) of the difference between young and elderly subjects for the total power (in g^2) in single axis spectra in the bandwidth ranges 0.25–2 Hz (left columns), 0.25–3 Hz (middle columns) and 0.5–11 Hz (right columns) during 6 tasks (rows; see legend of Fig. 3), derived from the power spectral density of the acceleration in axes x (upper panels), y (middle panels) and z (lower panels)

Task	Average ($\pm$ SD) total power in 0.25–2 Hz range			Average ($\pm$ SD) total power in 0.25–3 Hz range			Average ($\pm$ SD) total power in 0.5–11 Hz range		
	Young	Aged	P	Young	Aged	P	Young	Aged	P
x	FTT	0.005 $\pm$ 0.006		0.014 $\pm$ 0.013	0.014 $\pm$ 0.014	<	0.018 $\pm$ 0.020	0.029 $\pm$ 0.035	
	DDK	0.085 $\pm$ 0.124		0.172 $\pm$ 0.197	0.305 $\pm$ 0.253	<	1.602 $\pm$ 0.592	0.800 $\pm$ 0.457	<
	PIN	0.025 $\pm$ 0.013		0.017 $\pm$ 0.007	0.023 $\pm$ 0.010		0.064 $\pm$ 0.021	0.041 $\pm$ 0.019	<
	BRU	0.039 $\pm$ 0.030		0.025 $\pm$ 0.018	0.034 $\pm$ 0.025		0.243 $\pm$ 0.198	0.137 $\pm$ 0.084	<
	TXT	0.002 $\pm$ 0.001		0.004 $\pm$ 0.003	0.003 $\pm$ 0.001	0.01	0.008 $\pm$ 0.004	0.006 $\pm$ 0.003	0.10
	WALK	0.052 $\pm$ 0.016		0.040 $\pm$ 0.021	0.005 $\pm$ 0.004	>	0.070 $\pm$ 0.032	0.051 $\pm$ 0.027	
y	FTT	0.005 $\pm$ 0.011		0.018 $\pm$ 0.021	0.019 $\pm$ 0.021	>	0.006 $\pm$ 0.007	0.011 $\pm$ 0.012	
	DDK	0.023 $\pm$ 0.034		0.025 $\pm$ 0.030	0.035 $\pm$ 0.032		0.091 $\pm$ 0.082	0.080 $\pm$ 0.045	
	PIN	0.015 $\pm$ 0.010		0.023 $\pm$ 0.013	0.028 $\pm$ 0.016		0.027 $\pm$ 0.015	0.028 $\pm$ 0.015	
	BRU	0.014 $\pm$ 0.010		0.022 $\pm$ 0.016	0.031 $\pm$ 0.017	>	0.179 $\pm$ 0.139	0.165 $\pm$ 0.126	
	TXT	0.002 $\pm$ 0.002		0.004 $\pm$ 0.004	0.002 $\pm$ 0.003	>	0.003 $\pm$ 0.003	0.003 $\pm$ 0.002	
	WALK	0.103 $\pm$ 0.050		0.054 $\pm$ 0.047	0.059 $\pm$ 0.050	<	0.101 $\pm$ 0.032	0.047 $\pm$ 0.031	<
z	FTT	0.007 $\pm$ 0.013		0.026 $\pm$ 0.046	0.027 $\pm$ 0.047	>	0.016 $\pm$ 0.020	0.026 $\pm$ 0.024	
	DDK	0.095 $\pm$ 0.064		0.232 $\pm$ 0.221	0.337 $\pm$ 0.226		0.898 $\pm$ 0.522	0.564 $\pm$ 0.216	<
	PIN	0.021 $\pm$ 0.012		0.023 $\pm$ 0.014	0.032 $\pm$ 0.018		0.054 $\pm$ 0.024	0.038 $\pm$ 0.019	<
	BRU	0.028 $\pm$ 0.017		0.042 $\pm$ 0.038	0.050 $\pm$ 0.037		0.226 $\pm$ 0.147	0.119 $\pm$ 0.119	<
	TXT	0.006 $\pm$ 0.002		0.008 $\pm$ 0.006	0.007 $\pm$ 0.002		0.010 $\pm$ 0.002	0.006 $\pm$ 0.002	<
	WALK	0.046 $\pm$ 0.022		0.033 $\pm$ 0.019	0.037 $\pm$ 0.023		0.041 $\pm$ 0.019	0.034 $\pm$ 0.021	<

Only significance levels up to 0.10 are shown. Note that all significant differences and trends indicate a lower frequency and power in elderly subjects. Note that the significant effects and trends in the bandwidth range 0.5–11 Hz invariably show an age-related decline in power, whereas in the bandwidth ranges of 0.25–2 Hz and 0.25–3 Hz the gravitational artefact and bias for low-frequency movements conceal this effect, or may even cause a significantly higher power in elderly subjects. Direction of significant effects and trends is marked (< or >) to facilitate reading of the table.

4. Discussion

The present study shows that the acceleration of wrist movements is not limited to very low frequencies, as had been reported previously (Redmond and Hegge, 1985). The low-frequency components that dominated the frequency spectra reported by Redmond and Hegge (1985) probably result from differences in the position of the sensor in the gravitational field at the start and end of the time interval on which the Fourier transform has been applied. In the present study it has been shown that peak frequencies show substantial variability over subjects and activity types in the range of 0.25–11 Hz. Secondly, movement accelerations in elderly subjects are generally of lower frequency and intensity as compared to young subjects. Thirdly, the bandpass filter range of commercially available actigraphs is not optimal for studying wrist movements in general, and aging effects in particular. With a highpass setting of 0.25 Hz, the resulting activity counts are strongly affected by positional changes of the sensor in the gravitational field. Although it may be argued that changes in orientation of the accelerometer in the gravitational field can never occur without actual movement, the 'time above threshold', 'threshold crossing' and 'average acceleration' output variables of the actigraph do not properly measure acceleration induced by muscle force. A lowpass setting of 2–3 Hz will furthermore result in an underestimation of activity, and in a positive bias for actigraphic recordings in elderly subjects, as the frequency of their movement accelerations is in general lower than the movement acceleration frequency of young subjects. This bias may have contributed to the contradictory results in human aging studies using actigraphy, although this cannot be verified easily as most reports do not specify the frequency characteristics of the used actigraph. Actigraphs using single axis acceleration sensors will gain validity when the highpass filter is increased from 0.25 to 0.5 Hz, thereby reducing gravitational artefact, and when the low-pass filter is increased from 2 or 3 Hz to approximately 11 Hz. We are presently studying age effects on wrist activity with prototypes of a new actigraph that fulfils these requirements, and records both time above threshold and average acceleration (Van Someren et al., 1993).

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References

- Aschoff, J. (1994) On the aging of circadian systems. In: T. Hiroshige and K. Honma (Eds.), *Evolution of Circadian Clock*, Hokkaido University Press, Sapporo, pp. 23–44.
- Frost, J.D., Jr. (1978) Triaxial vector accelerometry: a method for quantifying tremor and ataxia, *IEEE Trans. Biomed. Eng.*, 25: 17–27.
- Lieberman, H.R., Wurtman, J.J. and Teicher, M.H. (1989) Circadian rhythms of activity in healthy young and elderly humans, *Neurobiol. Aging*, 10: 259–265.
- Redmond, D.P. and Hegge, F.W. (1985) Observations on the design and specification of a wrist-worn activity monitoring system, *Behav. Res. Meth. Instr. Comp.*, 17: 659–669.
- Renfrew, J.W., Pettigrew, K.D. and Rapoport, S.I. (1987) Motor activity and sleep duration as a function of age in healthy men, *Physiol. Behav.*, 41: 627–634.
- Ruchkin, D.S. and Glaser, E.M. (1976) Simple digital filters for examining CNV and P300 on a single-trial basis. In: D.A. Otto (Ed.), *Multidisciplinary Perspectives in Event Related Brain Potential Research*. Proc. 4th International Congress on Event-Related Slow Potential of the Brain (EPIC IV), Hendersonville, NC, pp. 579–605.
- Stone, P.A., Allen, M.T., Rayborn, G.H., and Harsh, J.R. (1992) The effects of nonstationarity, detrending and series length in Fourier analysis: a study with simulated data, *Sleep Res.*, 21: 358.
- Teulings, H.L. and Maarse, J.J. (1984) Digital recording and processing of handwriting movements, *Hum. Mov. Sci.*, 3: 193–217.
- Tryon, W.W. (1991) *Activity Measurement in Psychology and Medicine*, Plenum Press, New York, 247 pp.
- Van Hilten, J.J., Braat, E.A.M., Van der Velde, E.A., Middelkoop, H.A.M., Van Dijk, J.G., Ligthart, G.J. and Roos, R.A.C. (1995) Hypokinesia in Parkinson's disease: influence of age, disease severity, and disease duration, *Mov. Disorders*, in press.
- Van Someren, E.J.W., van Gool, W.A., Vonk, B.F.M., Mirmiran, M., Speelman, J.D., Bosch, D.A. and Swaab, D.F. (1993) Ambulatory monitoring of tremor and other movements before and after thalamotomy: a new quantitative technique, *J. Neurol. Sci.*, 117: 16–23.
- Van Someren, E.J.W., Hagebeuk, E.E.O., Lijzenga, C., Scheltens, P., De Rooij, S.E.J.A., Jonker, C., Pot, A.-M., Mirmiran, M. and Swaab, D.F. (1995) Circadian rest-activity rhythm disturbances in Alzheimer's disease, *Biol. Psychiat.*, in press.
- Witting, W., Kwa, I.H., Eikelenboom, P., Mirmiran, M. and Swaab, D.F. (1990) Alterations in the circadian rest-activity rhythm in aging and Alzheimer's disease, *Biol. Psychiat.*, 27: 563–572.