Technical note

Recording accelerations in body movements

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1 Introduction

OBJECTIVE MEASURES of the mobility of geriatric patients are necessary for better management of rehabilitation. At the moment there is a wide gap between the standard subjective clinical assessment and the full analysis practised by gait laboratories. Furthermore, complete gait analysis requires a dedicated laboratory and is impossible in a geriatric ward, let alone in the patient's home environment.

GUIMARAES and ISAACS (1980) showed that the gait of fallers had the following characteristics: slow speed, short step length, narrow stride width, wide range of stepping frequency, large variability of step length and increasing variability with increasing frequency. Assessment of these parameters in the patient's usual environment might be a prediction of falling, but no simple method of making such measurements exists. Methods of recording foot movement were reviewed by CRAWFORD *et al.* (1985). They either require foot-mounted contact switches with distance transducers, which by their very presence tend to affect what they are intended to measure, or recording walkways based on switches or on resistance changes, which limits their use to a laboratory environment. Accelerometry has often been attempted in gait analysis, usually without success, but MORRIS (1973) described its use in movements of the lower leg and reviewed the advantages and potential hazards of using accelerometers.

Low-cost, solid-state, piezoresistive accelerometers have recently become available and this note describes their incorporation into a portable instrument which records accelerations in three dimensions over a period of 1 min. The instrument has initially been used to monitor normal gait, but it is intended to use it in the study of the mobility of geriatric patients.

2 Methods

Movements in all three dimensions need to be recorded for the successful monitoring of mobility. Consideration of the gait cycle and walking speed of the elderly suggest that digitisation at 100 Hz for 1 min is sufficient to capture reliable information.

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A recording of the accelerations measured at the sacrum (Fig. 1) allows the identification of heelstrike in each foot and therefore the measurement of the time taken for each stride. If the person has walked normally over a path of known length, the data analysis gives the gait cycle, the cadence, the mean walking speed and the mean step length. Variations in the gait cycle are also easily measured as are differences between right and left steps.

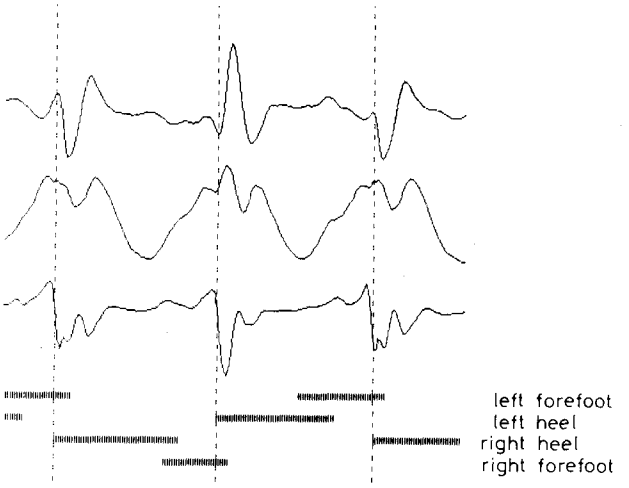

Fig. 1 *Three orthogonal acceleration signals from a normal healthy subject walking at a normal speed. Channel 1 (top) shows the lateral acceleration, channel 2 the up-down and channel 3 the anteroposterior signal. The fourth channel (lowest) is a signal from microswitches to show when the subject's heels and forefeet are in contact with the ground*

The reliability of the instrument was assessed on 20 normal, healthy volunteers. Recordings were taken of each subject walking at normal pace along a 20 m path. Each subject was asked to start the walk with a right step. The analysis of the data was performed by one of two investigators (GD or ALE). To reduce interobserver variation an agreed protocol was followed. The start of the walk was defined as the first step occurring simultaneously on all three channels. The end of the walk was defined as the last step appearing simultaneously on all three channels and occurring within the duration of two preceding gait cycles. To exclude variation in the subjects' gait at the beginning and end of the walk the first and last two steps were excluded in the calculation of temporal gait parameters. Ten recordings were analysed by both investigators to assess interobserver variation.

3 The instrument

The instrument (Fig. 2) is small and light and is held against the sacrum using a belt. The box has four protruding feet to discourage soft tissue movements. A removable memory card of capacity 32 kbytes protrudes slightly from the box.

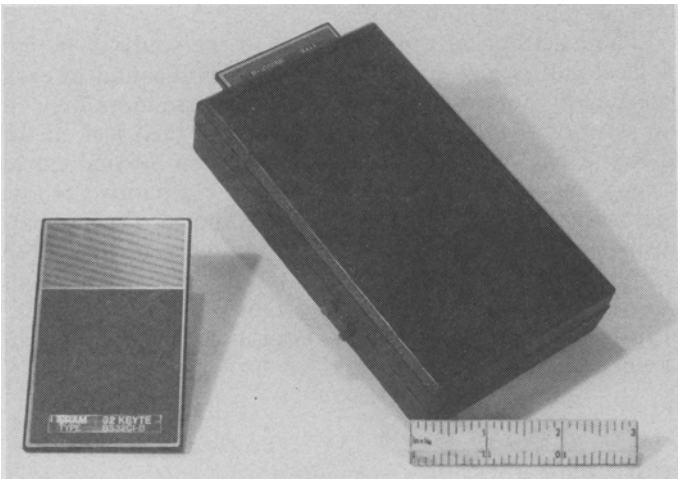

Fig. 2 *The instrument, which is small and light*

Once started, the instrument records the signals from three orthogonal accelerometers digitised to 8 bits at 140 Hz. The system's oscillator controls the data acquisition and the storage of the data in successive memory locations (Fig. 3). This is done by incrementing a 16-bit counter which directly addresses the memory and also provides the channel information for the analogue-to-digital convertor's multiplexer. When the 16-bit counter overflows, it stops the counter and the memory retains the recorded data. The memory card may then be removed and the data transferred to a personal computer for analysis.

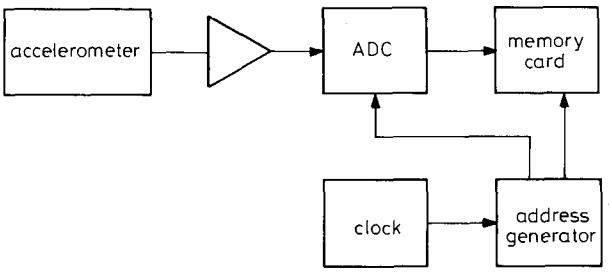

Fig. 3 *Block diagram of the instrument's electronic circuit*

The analysis programs display the data one screen at a time and allow the observer to identify and mark each heelstrike. The computer then calculates the time taken for each step and hence the mean value of the gait cycle and the cadence.

If the length of the path walked is known, the mean walking speed can be estimated as can the mean step length.

4 Results

The accelerometers and amplifiers respond to constant accelerations and therefore can be calibrated by orientating the box in different directions. Differences in acceleration values of $2g$ can therefore be produced and measured. The amplifier gains were selected so that an acceleration of $1g$ gave a 1 V signal or 20 per cent of the analogue-to-digital convertor's dynamic range. The calibration was also checked dynamically by dropping the instrument from a height of about 1 m into a box full of expanded polystyrene chips.

Fig. 1 shows the accelerations recorded by a young person with a normal gait. Each step is clearly defined and left and right steps can be distinguished by inspecting the direction of the lateral acceleration. For the purposes of demonstration and confirmation a spare analogue channel has been used to simultaneously record signals from four microswitches placed on the soles and heels of the subject's shoes. Heelstrike is most clearly seen in the antero-posterior channel but it can also be recognised in the pattern of channel 2, which represents acceleration in the vertical direction. Note that the inflections in the vertical channel signal can be used to identify the beginning of the step (heelstrike) and the end of the previous step (push-off),

Table 1 Gait parameters calculated from the accelerometer outputs for the 20 normal subjects

Males	Walking speed, $m\,s^{-1}$	Gait cycle, s	Cadence, min^{-1}	Step length, m	Right step time, s	Left step time, s
1	1·32	1·055	113·8	0·71	0·525	0·531
2	1·17	1·100	109·1	0·67	0·554	0·547
3	1·25	1·184	101·4	0·77	0·603	0·582
4	1·73	0·976	123·0	0·87	0·486	0·490
5	1·57	1·017	118·0	0·80	0·499	0·521
6	1·06	1·143	105·0	0·61	0·566	0·578
7	0·94	1·268	94·7	0·63	0·640	0·630
8	1·11	1·153	104·1	0·67	0·584	0·571
9	1·36	1·035	116·0	0·71	0·521	0·515
10	0·99	1·150	104·4	0·59	0·566	0·585
11	1·44	1·099	109·2	0·83	0·551	0·550
Mean ± SD	1·26 ± 0·24	1·107 ± 0·080	109·0 ± 7·8	0·715 ± 0·088	0·554 ± 0·043	0·554 ± 0·038
Females	Walking speed, $m\,s^{-1}$	Gait cycle, s	Cadence, min^{-1}	Step length m	Right step time, s	Left step time, s
1	1·33	0·991	121·1	0·67	0·511	0·486
2	0·88	1·274	94·2	0·59	0·657	0·620
3	1·35	0·955	125·6	0·67	0·495	0·465
4	1·31	1·049	114·4	0·69	0·529	0·521
5	1·45	0·969	123·9	0·71	0·465	0·506
6	1·55	0·908	132·1	0·71	0·462	0·442
7	1·86	0·917	130·9	0·87	0·462	0·454
8	1·56	0·913	131·5	0·71	0·466	0·446
9	1·44	1·009	118·9	0·74	0·510	0·500
Mean ± SD	1·41 ± 0·25	0·998 ± 0·107	121·4 ± 11·2	0·707 ± 0·070	0·506 ± 0·508	0·493 ± 0·052

thus giving an identifiable means of measuring the double-support and single-support times.

Table 1 shows the results of the 'gait analysis' for the 20 normals walking at their free walking speed. These results are comparable with those reported by SKINNER *et al.* (1988)

Interobserver variation in the calculation of these parameters was of the order of 2 per cent.

5 Discussion

MORRIS (1973) has described how many bioengineers have at some time attempted to use an accelerometer to study human movement. These attempts have usually failed because of the unsuitability of the accelerometer, either because of its poor low-frequency response or because of its excessive bulk and prohibitive price. The current work has used inexpensive, miniature, solid-state, piezoresistive accelerometers (IC Sensors Model 3021) which have a frequency response of 0–350 Hz.

A well recognised difficulty with the use of accelerometers in gait analysis has been the noise problem when acceleration is measured and then integrated to determine velocity and position. This work does not attempt this task; gait parameters are derived from measurements of time on the acceleration traces and mean values of walking speed are calculated from the total walking path length. It is suggested that the advantages of portability and universal application environment may in some cases outweigh the detailed position and spatial information which is given by the kinephotographic gait laboratory and to a lesser extent by instrumented walkways.

The results presented in Fig. 1 and Table 1 show that measurement of velocity, gait cycle, cadence and average step length are successfully performed by the system. Any asymmetry in the step times is also quickly apparent and may be quantified. Estimates of single-support and double-support times may also be performed, but these measurements require interpretation of the acceleration signals and are not reported in this note.

The extension of this technique to the study of pathological gait is now in progress. The identification of each step, easily done in normal gait, is of course more difficult in slowly moving patients who do not lift their feet off the floor. However, with care, identification of individual 'steps' has proved possible even in elderly patients whose gait might be described as a shuffle and also in patients using walking aids. Interesting results have already been obtained from hemiplegic patients and from Parkinson's disease patients. The technique promises to allow quantification of elderly patients' gait both in a hospital environment and in their own homes.

References

CRAWFORD, R. J., SANDFORD, M. H. and GILLESPIE, G. T. (1985) Automatic measurement of walking speed and stride length. *Eng. in Med.*, **14**, 101–106.

GUIMARAES, R. M. and ISAACS, B. (1980) Characteristics of the gait in old people who fall. *Int. Rehab. Med.*, **2**, 177–180.

MORRIS, J. R. W. (1973) Accelerometetry—a technique for the measurement of human body movements. *J. Biomech.*, **6**, 729–736.

SKINNER, S. B., SKINNER, H. B., and WYATT, M. P. (1988) Gait analysis. In *Encyclopaedia of medical devices and instrumentation.* WEBSTER, J. G. (Ed.), Wiley, New York, 1353–1365.

Technical note

Method for producing gas bubbles for use in air-embolism studies

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1 Introduction

INTRAVASCULAR GAS bubbles may occur during several surgical procedures (GOMAR *et al.*, 1985; MARSHALL and BEDFORD, 1981), or during infusion of fluids via central venous catheters (LAMBERT, 1982). Intravascular gas bubbles may also be formed during decompression and are believed to be an important aetiological factor in the development of decompression sickness (DCS). Formation of such gas bubbles is primarily located to the venous system, and it is widely accepted that the pulmonary system acts as a filter and traps venous gas bubbles (EMERSON *et al.*, 1967). More recent experiments have, however, demonstrated that gas bubbles may escape lung entrapment and enter the arterial circulation (BUTLER and HILLS, 1979; 1981; 1985; BUTLER and KATZ, 1988). Arterial gas bubbles that enter the carotid artery may cause dysfunction in the brain (CURLEY *et al.*, 1988; DUTKA *et al.*, 1988) or interfere with the blood/brain barrier (CHRYSSANTHOU *et al.*, 1987).

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