A microcomputer-based system for measuring temporal asymmetry in amputee gait.

C. CHEUNG, J. C. WALL, and S. ZELIN.

School of Engineering, University of Guelph, Ontario, Canada.

Abstract
This paper reports the development of a microcomputer-based system for measuring the temporal factors of amputee gait. The system has been designed for possible use in a standard clinical environment. It consists of a simple walkway, two photoelectric beam relays, a control unit and an APPLE II Plus microcomputer.

Preliminary clinical assessment of the system has been conducted. The gait characteristics of eight amputees during their ambulation training were investigated with the use of the new system. A comparison of the results from different groups of subjects, selected primarily on level of amputation, is presented.

Introduction
Human gait is an intricate combination of the neuromuscular and skeletal systems working together to produce a smooth and efficient form of locomotion. In upright bipedality, man's normal mode of locomotion, the lower limbs are the prime structures which not only support the body but provide power and control. The lower limb amputee has been deprived of part of his natural locomotor system and, in order to restore some functional mobiliy, he must learn to walk on an artificial limb. Thus one of the major goals in rehabilitating the amputee is to improve his ability to walk on his prosthesis. To monitor the progress made, if any, during the rehabilitation programme, assessment of the amputee's walking pattern must be made.

Subjective assessment techniques have been commonly used to evaluate gait. Although it is easy enough for a clinician to recognise major improvements in the walk of a patient, minor

All correspondence to be addressed to Dr. J. C. Wall, School of Physiotherapy, Dalhousie University, Halifax, Nova Scotia, Canada, B3H 3J5

changes are more difficult to detect. More importantly, recording accurately the manner in which the patient walks such that comparison of a patient's walking performance can be made with others of similar disability and age is a very difficult problem. Hence the need for objective measurements of gait becomes apparent.

Many techniques have been developed to measure human gait objectively. These gait assessment methods attempt to measure the following parameters: temporal and distance factors, force analysis, angular displacements of body segments, electromyographic activity, and metabolic cost (Wall et al, 1976). The measurement of all these parameters simultaneously would be necessary for a reasonably complete analysis of gait and would be ideal for research purposes. However, it would provide an unwieldy amount of data for clinical purposes. For clinical applications, it is desirable to provide a simple objective measurement of some parameters of gait which could aid in the assessment of the quality of gait. Obviously, the parameters have to be chosen according to the pathology being studied.

Many investigators have included the measurement of the temporal factors of gait in their studies of amputees (Zuniga et al, 1972). These temporal factors include stride time, support time, swing time, double support time, and their derivatives. Table 1 is used to summarize the results for the temporal and distance measurements from these studies. These results indicate that one of the gait characteristics of amputees is an asymmetrical walking pattern. The asymmetry can be seen in the measurements of step lengths and the various temporal factors. It seems reasonable to hypothesize that "good" gait is more symmetrical than "bad" gait, and it is suggested that the degree of symmetry should increase as

one's walking ability improves. Therefore, the quality of amputee gait may be indicated by the degree of symmetry in the measurements of step lengths or temporal factors.

The differences in gait between amputees and normals are ones which appear to provide a sense of safety, minimize pain and compensate for the problems related to walking with a prosthetic limb. Due to these anomalies, the amputee is walking at a higher metabolic cost than normal. The level of amputation has proven to be important in dictating the severity of the gait deviations.

The data shown in Table 1 are obtained mostly from studies on amputees who have used their prostheses for a considerable period of time, and who have been judged clinically to have good gait patterns. In addition, the age of these amputees averaged about 45. Therefore, it is not unreasonable to presume that the deviations would be more pronounced for new amputees

during the gait training period. It is felt that objective measurement of gait during this period can be of great benefit to the therapist in assessing and treating the amputee. Such measurement may also be of assistance to the prosthetist in making decisions regarding modifications to the artificial limb.

Since the temporal factors appear to reflect the anomalies of the amputee's gait and these parameters are relatively simple to measure, it was decided to develop a system for their measurement. Various techniques have been developed to determine these temporal factors. Winter et al, (1972) have developed micro-switches for affixing to the subject's shoes. Drillis (1958) and Gardner and Murray (1975) have used metal walkways with conductive strips on the subject's shoes. Special instrumented mats have been developed by Wall et al, (1976), Gabel et al, (1979) and Taylor (1980). Photographic techniques have been used by

Table 1. Temporal and distance factors for normal individuals, above-knee amputees and below-knee amputees.

Parameters	Normal Individuals	Unilateral AK Amputees		Unilateral BK Amputees	
Preferred walking speed (m/s)	1.51	1.00 [2] 1.03 [3] 0.94 [4]		1.07 [5]	
Cadence (steps/min)	113 [1]	87 [2] 91 [3] 85 [4]		96 [3]	
Stride Time (s)	1.06 [1]	1.38 [2] 1.41 [4] 1.43 [6]		1.26 [5]	
		Prosthetic	Anatomic	Prosthetic	Anatomic
Step Length (m)	0.78 [1]	0.72 0.68	0.64 [2] 0.62 [4]	0.68	0.63 [5]
Support Time (% stride time)	61 [1]	58 57 58	68 [2] 65 [4] 66 [6]	61	65 [5]
Swing Time (% stride time)	39 [1]	42 43 42	32 [2] 35 [4] 34 [6]	39	35 [5]
Double Braking Support Time (% stride time)	11 [1]	14 11 13	12 [2] 11 [4] 10 [6]	—	—
Swing Support	0.63 [1] 0.62 [6]	0.72 0.75 0.72	0.47 [2] 0.54 [4] 0.52 [6]	0.64	0.54 [5]

[1] Murray et al, (1964)
[2] Murray et al, (1980)
[3] Drillis (1958)

[4] James and Oberg (1973)
[5] Robinson et al, (1977)
[6] Zuniga et al, (1972)

Murray et al, (1964). The feasibility of employing any system in a clinical setting must be evaluated in terms of cost, simplicity of construction, portability, simplicity of use, encumbrance to the patient, time taken to obtain the results, the manner of result presentation, and finally, speed and accuracy (Wall et al, 1980).

The advent of relatively low cost and more readily available microcomputers has opened up new possibilities in the area of gait assessment. It is felt that incorporating general purpose microcomputers with the existing gait measurement techniques will provide systems which have the potential for clinical applications. This paper describes the development of one such system designed to measure temporal asymmetry of amputee gait.

Methods

The system consists of a walkway, two sets of photoelectric beam relays, a control unit and an APPLE II Plus microcomputer. The walkway is designed to provide an electrical circuit which can detect the moments of double limb support during the subject's walk. To provide both portability and flexibilty, a modular approach has been taken in constructing the walkway. It is made up of four individual sections. Each of the mid-sections consists of two sheets of aluminium

foil (1.22m × 0.45m) glued to a piece of pressboard (1.22m × 0.91m), leaving a gap of 0.5cm in the centre. The end sections are made in a similar fashion, except that a non-conductive surface of 0.44m in length is left at one end. This surface is used to allow the subject to stand on the walkway prior to each test session without interfering with the electrical circuit of the walkway. The four sections are connected by metal connection plates to form a walkway 4.88m in length. To provide a non-slip walking surface the aluminium surface is coated with a black conductive rubber paint. The remaining portions of the walkway are painted with a non-conductive rubber paint to provide a uniform appearance. The walkway is shown schematically in Figure 1. It should be noted that the walkway can be shortened or lengthened by removing or adding sections to the middle of the walkway. The dimensions given above were dictated by the parallel bars between which the walkway was designed to be used.

One side of the walkway is connected to a 5v DC power supply and the other side is connected to the input of a Schmitt trigger circuit. Each of the conductive surfaces of the walkway serves as one contact of a switch. Each piece of the self-adhesive aluminium tape affixed to the subject's shoes serves as the other contact of the respective switch. Electrical continuity between the two pieces of tape is provided by clipping a thin wire to the tape. This wire runs from one foot to the other. Placement of the foot on the metal surface of the walkway closes the
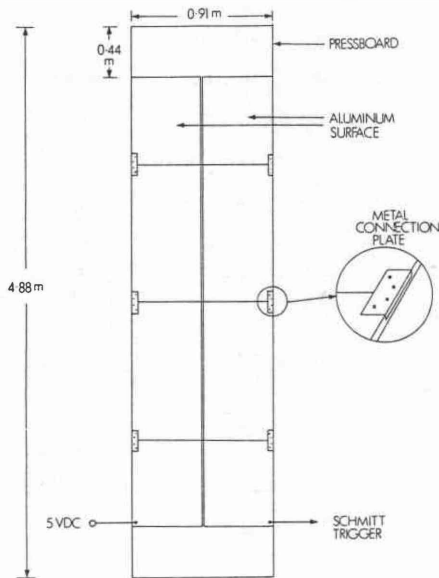

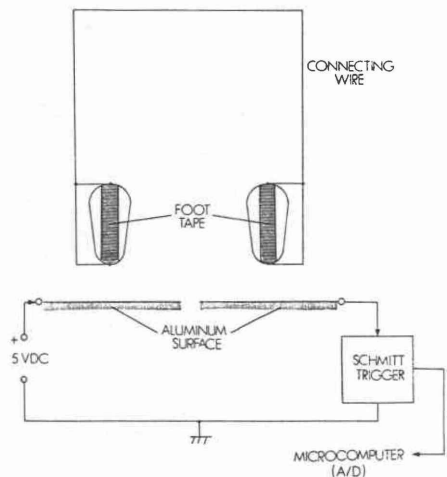
Fig. 1. Schematic representation of walkway.

Fig 2. Basic circuitry of the walkway.

respective switch. The basic circuit of the walkway is shown in Figure 2.

If the subject stands on one side of the walkway with one foot in contact, the circuit remains open until the second foot comes into contact with the other side of the walkway. The circuit then closes and a 5v signal appears across the input part of the Schmitt trigger. The circuit remains closed during the time interval when both feet are in contact (i.e. double support phase). At the moment either foot is removed from the walkway and swings forward (swing phase or single support phase on the contralateral side), the circuit is re-opened and the input signal to the Schmitt trigger drops to zero. Figure 3 shows diagrammatically the level of the Schmitt trigger output at different phases of the gait cycle. Thus, while the subject is walking, the output signal changes state in an alternating fashion at the onset, and at the end, of each double support phase. By monitoring the status of the signal, the temporal phase of gait can be resolved.

The system also employs two sets of photoelectric beam relays to determine the time required by the subject to tansverse a fixed distance, hence the average walking speed. The outlet from each relay is connected to the input of a Set-Reset Flip-Flop circuit. During operation of the system, the two relays are placed at shoulder level, four metres apart on the walkway. When the subject enters the central four metre region of the walkway, he/she will interrupt the light beam of the first relay which causes the normally-open relay contact to close. Closure of the contact will set the Flip-Flop output. The output remains set until the subject interrupts the second light beam at the end of the four metre region. The output is then reset. By determining the duration that the output remained set, the time for the four metre excursion can be resolved.

To enable measurements to be taken when walking is initiated from either end of the walkway, the outlets of the relays are connected to a double-pole-double-throw switch, thus the same relay can set or reset the Flip-Flop circuit depending on the position of the switch. Two pushbutton switches are also included in the circuit to make manual set/reset of the circuit possible.

To collect, store and process the data from the walkway and relays, an APPLE II Plus micrcomputer is employed. The microcomputer has 48k of random access memory, an on-board analog-to-digital converter and an on-board real
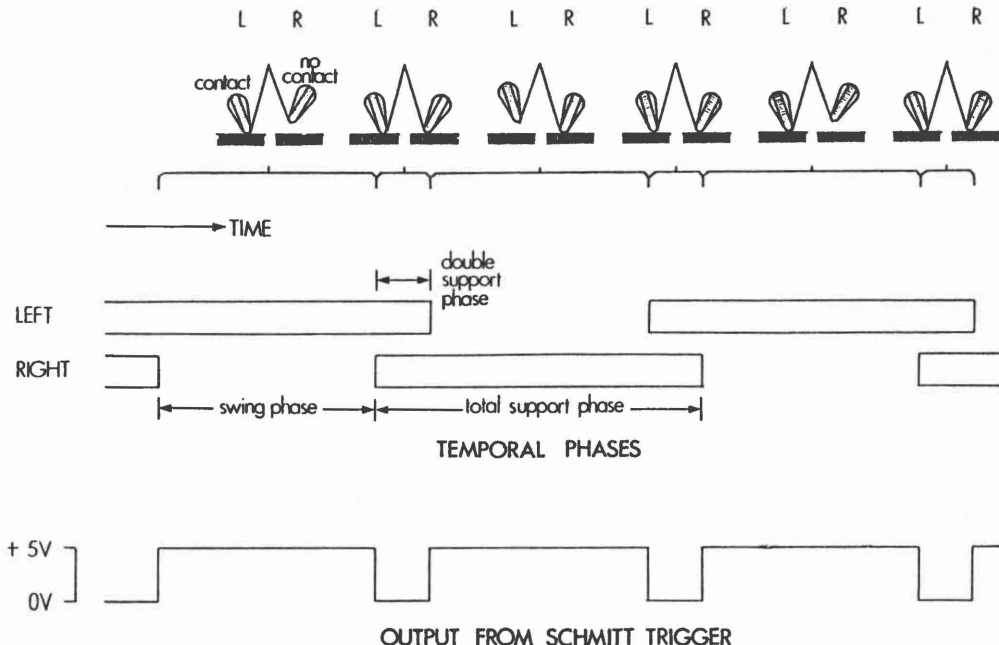

Fig. 3. Schematic representation of the level of the Schmitt trigger output at different phases of the gait cycle.

time clock. Other peripherals include a floppy disk drive, a thermal printer and a video monitor.

A control unit has been built to provide an interface between the microcomputer and the other hardware. It contains the Schmitt trigger and the Flip-Flop circuit. A 5v DC power supply is also built into the control unit to provide the voltage required by all the electronic components. LED and control switches are also mounted on the control unit. The unit also provides an interface for the output signals to the microcomputer and to any strip chart recorder. The output signals are fed into two separate channels of the analog-to-digital converter in the microcomputer for on-line measurement. A block diagram of the system is shown in Figure 4.

During a test session, the microcomputer uses a machine language routine to sample the incoming signals. This routine monitors the voltage level of the two signals alternately. When the signal is found to have changed state, the routine reads the time from the real time clock and stores the time readings in a buffer region set aside in the random-access memory. The routine also records the number of times that the Schmitt trigger signal changed state. This recording procedure will be terminated when the Flip-Flop is found to be reset (i.e. the subject has interrupted the second beam). Because the two signals are examined alternately, a delay may occur between the moment that a signal changes state and the detection of that change. The magnitude of this delay depends upon the

execution time of the machine language instructions, the time required to digitize the signal, and the time required to access and record time readings from the clock. The maximum delay, however, is estimated to be 4 ms which is acceptable and well within the 10 ms resolution required.

Once the recording procedure is terminated, the main program retrieves the time readings from the buffer region and determines the various temporal factors and velocity. To determine the temporal factors, the operator has to indicate which foot, left or right, came into contact when the signal first changed state. This information is important in determining the proper sequence of foot contacts. Once the results are determined, the program then stores the results in a floppy disk file for future reference and prints the results.

The velocity is measured in the standard units of metres per second as well as in terms of relative speed. This method of normalizing walking speed to take into account differences in height, was suggested by Grieve and Gear (1966) and has been used in many studies since, including those by Charteris et al, 1981; Wall et al, 1980; Wall et al, 1981; and Taves et al, 1982. One of the reasons for the growing acceptance of this form of measurement is that the correlation between walking speed and the temporal measures such as stride time, is greater when speed is expressed in relative terms as opposed to the absolute terms of metres per second. (Rosenrot et al, 1980).
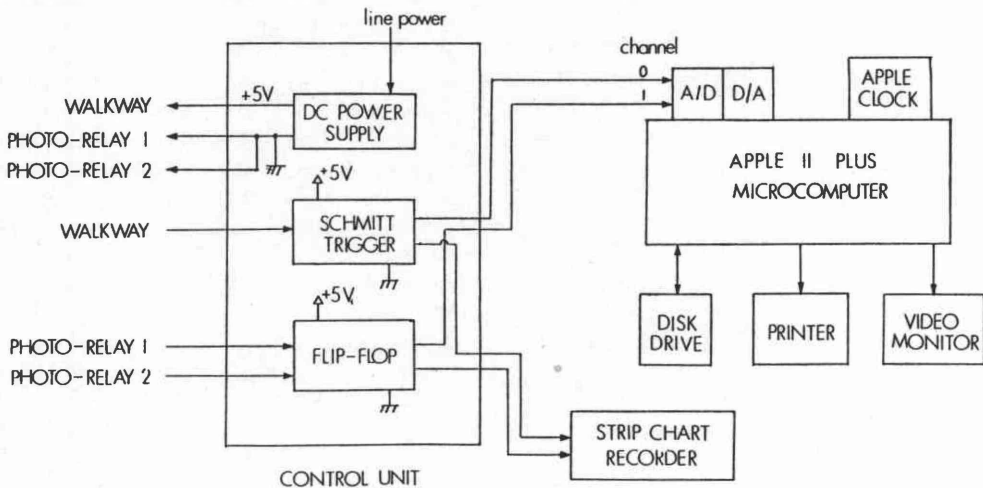

Fig. 4. Block diagram of the system.

The temporal parameters that can be determined from the measurement of the double support phases are shown in Figure 3. Single support time, which is not shown in this diagram, is equal to contralateral swing time. To distinguish the two double support phases that occur in a single stride the term "Double Braking Support" has been used. This term refers to the period of double support following heel strike. Thus left double braking support is the interval between left heel strike and right toe-off. This is the phase which Ducroquet et al, 1968 refer to as the "Double Anterior Support of Reception".

The result sheet is designed to present the temporal data both numerically and graphically in a from similar to that used by Wall et al, (1976). On the result sheet, the temporal factors for each stride cycle measured are given numerically in terms of raw time (seconds) and in percent of stride time. This permits a stride-to-stride comparison of the subject's walk. The mean values are also included. To aid in the interpretation of the data, two graphical plots are used. A real time factor graph is constructed from the temporal data as collected during the walk. It is used to indicate the times of contact of the subject's feet during the test, thus allowing a stride-to-stride comparison. A mean time factor graph is constructed by plotting the mean values of the temporal factors over a period of 5 seconds. This graph is designed to illustrate the average walking performance of the subject during one particular test, and to allow for easy comparison between different tests.

Subjects

The subjects who volunteered for this study were recent amputees undergoing a gait training

Table 2. Subject statistics

Subject No.	Sex	Age	Height (m)	Site of Amputation
1*	F	50	1.64	Right—AK
2	F	62	1.60	Right—AK
3	M	68	1.70	Left— BK
4	F	71	1.58	Right—BK
5	M	54	1.78	Right—BK
6	M	59	1.65	Right—BK
7**	M	49	1.67	Left— BK
8***	F	70	1.52	Left— BK Right—BK

* Subject 1 had left big toe amputated and also wore an orthosis on that leg.
** Subject 7 was blind.
*** Subject 8 had left leg amputated three years before the right leg amputated.

programme. Included were two unilateral above-knee (AK) amputees, five unilateral below-knee (BK) amputees and one bilateral below-knee amputee. Table 2 outlines the statistics for each subject. It should be noted that Subject 1 had the big toe of the anatomic leg amputated and also wore an orthosis on that leg during the period of training. In addition, Subject 7 is blind. These factors may have influenced the results.

Each of the AK amputees wore a prosthesis which had a quadrilateral total contact socket, a single axis safety knee component and a solid-ankle-cushion-heel (SACH) foot. The prosthesis was suspended by a Silesian bandage. All BK amputees wore a prosthesis which was suspended by a pelvic band and had a patellar-tendon-bearing socket and a SACH foot.

These subjects were tested on different occasions during the course of their gait training. On each occasion, the subject was tested on four walks after two practice walks. Results from these four walks were used to determine the average walking performance of the subject on that day.

Results

Since the primary aim of the clinical study was to assess the clinical suitability of the walkway system, it was decided to treat the data by group as far as possible. It is well-documented that the temporal gait characteristics of AK amputees are significantly different from those of BK amputees. The following analysis is an attempt to compare the gait patterns of these two groups during the gait training period. To this end, the two AK amputees (Subjects 1 and 2) in the study were considered as one group, and four BK amputees (Subjects 3, 4, 5 and 6) were chosen as one group. Although Subject 7 is also a BK amputee, his gait was found to be significantly different from the other BK amputees, possibly because of his blindness. Therefore, it was decided to consider him individually. Furthermore, Subject 8 is also being considered individually because of her bilateral amputations.

For ease of comparison, the training period has been divided into three stages. The period of training for the subjects in the study was approximately six weeks, thus, it was decided to divide the training period into three two-week periods. For each subject, the results obtained

Table 3. A comparison of gait characteristics between selected subject groups in different periods of training.

Group	Training Period (weeks)	Speed (stat./s)	Stride Time (s)	Total Support Time (% stride time)			Double Braking Support Time (% stride time)		
Above-knee				Pros.	Anat.	Asym.	Pros.	Anat.	Asym.
Amputees	0-2	0.09	4.09	75.7	87.5	-11.8	39.8	23.4	16.4
(Subjects	2-4	0.14	2.69	74.2	82.4	-8.2	33.5	23.0	10.5
1 and 2)	4-6	0.14	2.47	69.4	79.3	-9.9	28.2	20.8	7.4
Below-knee				Pros.	Anat.	Asym.	Pros.	Anat.	Asym.
Amputees	0-2	0.19	2.23	74.3	80.0	-5.7	30.5	24.3	6.2
(Subjects	2-4	0.21	2.11	74.1	78.9	-4.8	29.6	23.2	6.4
3, 4, 5, 6)	4-6	0.27	1.92	71.7	75.2	-3.5	27.6	20.2	7.4
Subject 7				Pros.	Anat.	Asym.	Pros.	Anat.	Asym.
(Blind	0-2	0.15	2.42	66.0	80.5	-14.5	33.2	13.8	19.4
Below-knee	2-4	0.21	2.08	62.8	77.9	-15.1	30.0	11.4	18.6
Amputee)	4-6	0.20	2.09	66.4	77.4	-11.0	27.5	17.1	10.4
Subject 8				Right	Left	Asym.	Right	Left	Asym.
(Bilateral	0-2	0.11	2.88	82.0	85.5	-3.5	40.5	26.2	14.3
Below-knee	2-4	0.15	2.41	79.2	82.7	-3.5	35.9	26.2	9.7
Amputee)	4-6	0.17	2.28	81.2	83.9	-2.7	37.6	26.8	10.8

Note— Pros = prosthetic side; Anat = anatomic side: Asym = asymmetry (prosthetic–anatomic, except for Subject 8 where asymmetry = right–left.)

within a specific period were used to determine his/her average performance for that period of time. As a group, the average performance within each specific period was computed from the corresponding subjects' average performance in that period of time.

The results for the four groups of subjects are shown in Table 3. The graph (Fig. 5) illustrating the walking speeds for these groups showed an improvement during the gait training period. The walking speed of the BK amputee group, by the second two-week period, and even more noticeably in the last stage, revealed a marked improvement. The AK amputee group showed a significant improvement from the first stage to the second stage, but no change from the second to the last stage. In all periods of training, the BK amputees displayed a higher walking speed than the AK amputees. The speed of subject 7 also showed a marked improvement initially, and, although his speed decreased in the last stage, the change was only 0.01 statures/s. In general, he walked faster than the AK amputees, but slightly slower than the other BK amputees. The bilateral BK subject displayed improvement in walking speed over the training period. In addition, she exhibited a slightly higher walking speed than the AK amputees.

Figure 6 shows a decrease in stride time for all groups over the training period. The AK amputee group showed the most dramatic change in stride time, especially from the first stage to the second stage of training. The other groups displayed a more gradual change in stride time. A strong correspondence between the walking speed and stride time has been found. For example, the walking speed of the AK amputee group was shown to be slower than that of the BK subjects, whereas the stride time of the AK amputee group was found to be longer than that of the BK amputee group. Similar relationships can be found in the other groups.
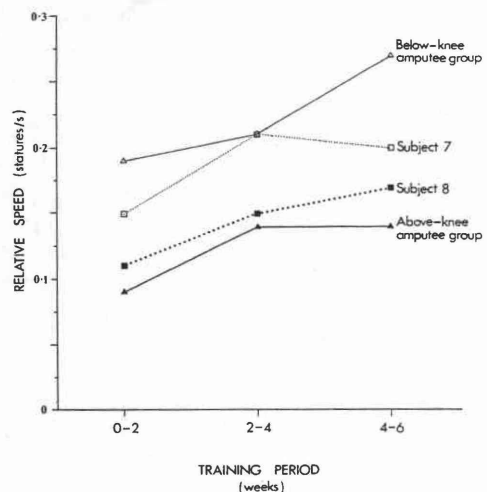

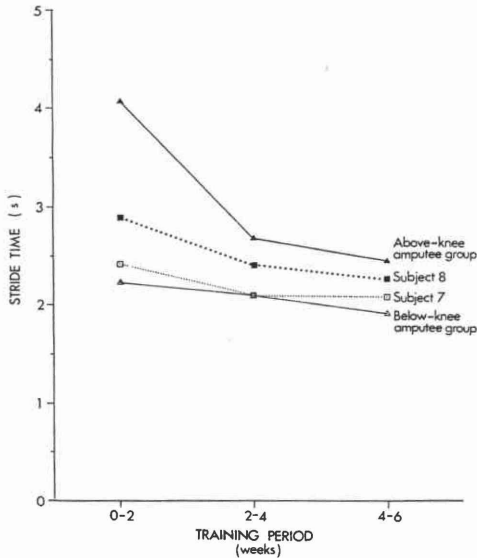Fig. 5. Comparison of walking speeds of selected subject groups in different periods of training.

Fig 6. Comparison of stride times of selected subject groups in different periods of training.

was longer than that on the right side. The AK amputee group showed a marked reduction in the degree of asymmetry, especially from the first period to the second period of training. The BK amputee group showed a slight increase in the degree of asymmetry during the course of training, however the degree of asymmetry was smaller than that of the AK amputee group. Again, Subject 7's gait revealed the highest degree of asymmetry when compared to the other groups. However, his asymmetry was shown to be greatly reduced during the course of training, especially by the last training period. In the first and second periods of training, the degree of asymmetry in double braking support times of Subject 8 was comparable to that of the AK amputee group. Improvement was also noted here. However, in the last period of training, there was a slight increase in the magnitude of her asymmetry.

Marked differences in the gait of AK and BK amputees during the course of training have been demonstrated in the above comparisons. The gait patterns of the two special subjects were found to be characteristically different from

For all groups, there are indications that the improved speed was achieved by decreasing stride time.

The quality of these subjects can be further demonstrated by considering the asymmetry in the temporal factors. Figure 7 illustrates the asymmetry in total support times and in double braking support times for all groups of subjects. Since the asymmetry in swing times is identical to that in total support times, it is not included in this comparison. Figure 7a indicates that for all subjects with unilateral amputation, the total support time of the anatomic side is longer than that of the prosthetic side. For the bilateral BK subject, the left total support time was found to be longer than the right total support time. The degree of asymmetry showed a reduction over the training period in all groups, and is an indication of an improvement in the quality of gait. Comparison of the performance of different groups showed that Subject 7 had the highest degree of asymmetry and Subject 8 had the lowest degree of asymmetry. In addition, the AK amputee group displayed a higher degree of asymmetry than the BK amputee group.

Figure 7b indicates that for all unilateral amputees, the double braking support time is longer on the prosthetic side than on the anatomic side. For the bilateral BK subject it was found that the left double braking support time
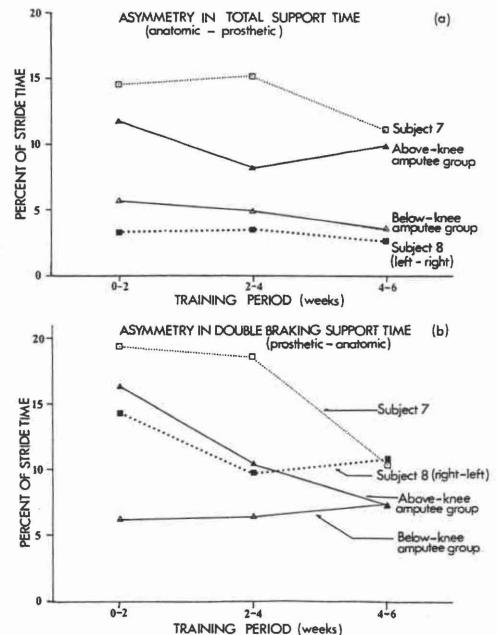

Fig 7. Comparison of (a) asymmetry in total support time and (b) asymmetry in double braking support time of selected subject groups in different periods of training.

those of the other groups. Subject 7, possibly because of his blindness, displayed a very asymmetrical gait. However, in terms of walking speed and stride time, his gait is comparable with the other BK amputee subjects. For the bilateral BK amputee subject, both her walking speed and stride time fall between those of the AK and BK amputee groups. She displayed a higher degree of asymmetry in the double braking support times than in the total support times.

Conclusion

Preliminary clinical assessment indicates that the system seems to be acceptable for clinical applications. The fact that the microcomputer is relatively small in size and the walkway is modular, allows the system to be set up in any standard physical therapy room. The walkway is flat and can easily be assembled and disassembled, allowing the system to be transported easily from one clinic to another. The speed and accuracy of the measurements provided by the system have been shown to be acceptable for the purposes intended. The technique used to collect the temporal data seems to be acceptable, from the clinician's viewpoint, because it produces a record of the patient's walking in a natural manner, unencumbered by apparatus and in the patient's normal shoes. The only attachment to the patient is a thin wire connecting the two strips of conducting tape which are placed on the soles of the patient's own shoes. A minor drawback of the system is that patients are required to walk with their feet on either side of the walkway's centre line. However, results from the clinical study seem to indicate this does not impose too great a constraint on the patients, particularly lower limb amputees who tend to have a wide base of support during walking.

Results from the clinical study have revealed the temporal gait patterns of eight amputee subjects during their gait training programme. As a group, the BK amputees displayed a higher walking speed than the AK amputees. Correspondingly, the stride time of the BK amputee group was shorter than that of the AK amputee group. In addition, the asymmetry in the temporal phase was found to be more pronounced in the AK amputee group than in the BK amputee group. The improvement shown in the AK amputees' gait over the training period, was more dramatic than that of the BK

amputees, particularly in the early stages of training. The two special subjects in the study exhibited very different gait patterns from the others. The reason may be related to the physical characteristics of these subjects. It should be emphasized here that the results from the study were obtained from the subjects walking with the use of the parallel bars for support. This may be an important consideration when one wishes to compare these results with those from other studies.

Because of the small number of subjects and the significant differences between these subjects, the results from this study should be considered tentative. It is not the intention of this study to generalize the gait patterns of amputees during gait training. However, the results from the study can be used to illustrate the usefulness of the system in monitoring the progress of ability during gait training, and in allowing for comparison of a patient's walk with others of similar pathology.

Acknowledgements

REFERENCES

CHARTERIS, J., WALL, J. C., NOTTRODT, J. W. (1981). Functional reconstruction of gait from the Pliocene hominid footprints at Laetoli, Northern Tanzania. *Nature*, **290**, 496–498.

DEWAR, M., NEIL, M. F. (1978). Amputee gait analysis and biofeedback. In: Department of Health and Social Security. *Biomechanical Research and Development Unit Report*, 245–251.

DRILLIS, R. J. (1958). Objective recording and bio-mechanics of pathological gait. *Ann. N.Y. Acad. Sci.*, **74**, 86–109.

DUCROQUET, R., DUCROQUET, J., DUCROQUET, P. (1968). *Walking and limping: a study of normal and pathological walking*. Philadelphia, P.A., Lippincott.

GABEL, R. H., JOHNSON, R. C., CROWNINSHIELD, R. D. (1979). A gait analyzer/trainer instrumentation system. *J. Biomech.*, **12**, 543–549.

GARDNER, G. M., MURRAY, M. P. (1975). A method of measuring the duration of foot-floor contact during walking. *Phys. Ther.*, **55**, 751–756.

GRIEVE, D. W., GEAR, R. J. (1966). The relationships between length of stride, step frequency, time of swing and speed of walking for children and adults. *Ergonomics*, **5**, 379–399.

JAMES, U., OBERG, K. (1973). Prosthetic gait pattern in unilateral above-knee amputees. *Scand. J. Rehabil. Med.*, **5**, 35–50.

MURRAY, M. P., DROUGHT, A. B., KORY, R. C. (1964). Walking patterns of normal men. *J. Bone Joint Surg.*, **46A**, 335–360.

MURRAY, M. P., SEPIC, S. B., GARDNER, G. M., MOLLINGER, L. A. (1980). Gait Patterns of above-knee amputees using constant-friction knee components. *Bull. Prosthet. Res.*, **17(2)**, 35–42.

PERRY, J., BONTRAGER, E., ANTONELLI, D. (1979). Foot switch definition of basic gait characteristics. In: Kenedi R. et al, (eds). *Disabilities: proceedings of a seminar on rehabilitation of the disabled*. London: Macmillan. p 131–135.

ROBINSON, J. L., SMIDT, G. L., ARORA, J. S. (1977). Accelerographic, temporal, and distance gait factors in below-knee amputees. *Phys. Ther.*, **57**, 898–904.

ROSENROT, P., WALL, J. C., CHARTERIS, J. (1980). The relationship between velocity, stride time, support time and swing time during normal walking. *J. Hum. Move. Stud.*, **6**, 323–335.

TAVES, C., CHARTERIS, J., WALL, J. C. (1982). The kinematics of treadmill walking during pregnancy. *Physiotheraphy Canada*, **34**, 321–324.

TAYLOR, D. R. (1980). An instrumented gait mat. In: *Proceedings of the International Conference on Rehabilitation Engineering. Toronto, Canada.* Ottowa, Ontario: Canadian Medical and Biological Engineering Society. p. 278–279.

WALL, J. C., DHANENDRAN, M., KLENERMAN, L. (1976). A method of measuring the temporal/distance factors of gait. *Biomed. Eng.*, **11**, 409–412.

WALL, J. C., ASHBURN, A., KLENERMAN, L. (1980). Gait analysis in rehabilitation medicine. *Proceedings of the International Conference on Rehabilitation Engineering. Toronto, Canada.* Ottowa, Ontario: Canadian Medical and Biological Engineering Society. p. 284–286.

WALL, J. C., ASHBURN, A., KLENERMAN, L. (1981). Gait analysis in the assessment of functional performance before and after total hip replacement. *J. Biomed. Eng.*, **3**, 121–127.

WINTER, D. A., GREENLAW, R. K., HOBSON, D. A. (1972). A microswitch shoe for use in locomotion studies. *J. Biomech.*, **5**, 553–554.

ZUNIGA, E. N., LEAVITT, L. A., CALVERT, J. C., CANZONERI, J., PETERSON, C. R. (1972). Gait patterns in above-knee amputees. *Arch. Phys. Med. Rehabili.*, **53**, 373–382.