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ANALYSIS OF HUMAN MOTION IN REHABILITATION BY MICRO-COMPUTER

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ABSTRACT

Analysis of human motion is extremely important in order to obtain fundamental data for training in the field of rehabilitation. At Nagoya University we are analysing human motion by use of a micro-computer to elucidate functional characteristics of the locomotion system. After the introduction of micro-computers, it has become possible to accumulate and integrate more data and process them more quickly and conveniently. Data obtained from the weight balance analyzer, large force plate and three-dimensional goniometer devised by the authors are subjected to on-line processing through an A/D convertor. 1. The weight-balance analyzer for the evaluation of dynamic balance ability can be used both in standing and sitting positions. A computer block-break game is integrated into the program so that training can be performed enjoyably. 2. A large serial force plate is used to detect the floor reaction force of individual legs. Three force components in the directions of X, Y and Z are transformed to resultant force, velocity, acceleration and paths of center of pressure through mathematic processing. 3. The three-dimensinal goniometer describes angular changes of the hip or knee joint in the three planes during walking. In addition to flexion-extension, adduction-abduction and internal-external rotation can be grasped easily by this device.

Keywords: Micro-computer, Weight balance analyzer, Floor reaction force, Three-dimensional goniometer, Rehabilitation medicine

INTRODUCTION

Motion analysis in rehabilitation is extremely important in order to obtain fundamental data for exercise therapy, and is clinically useful since it demonstrates human motion objectively, quantitatively, graphically and visually for a short time. Since there is a fear that such analysis is performed only for the purpose of analysis, as is the often case, care must be taken so that motion analysis is always performed for the purpose of treatment and training.

Micro-computers play an important role in the measurement and description of information obtained from various sensors. This paper deals with motion analysis in rehabilitation using a Mark III microcomputer (Sord Co., Ltd.) at the Nagoya University Hospital.

MATERIALS AND RESULTS

Weight Balance Analyzer (WBA)

1. *WBA as an evaluation device of equilibrium function and as a training machine*

This device was developed and put into practical use by the authors to quantitatively determine at varying times the ability of body-weight shifting in dynamic standing of the

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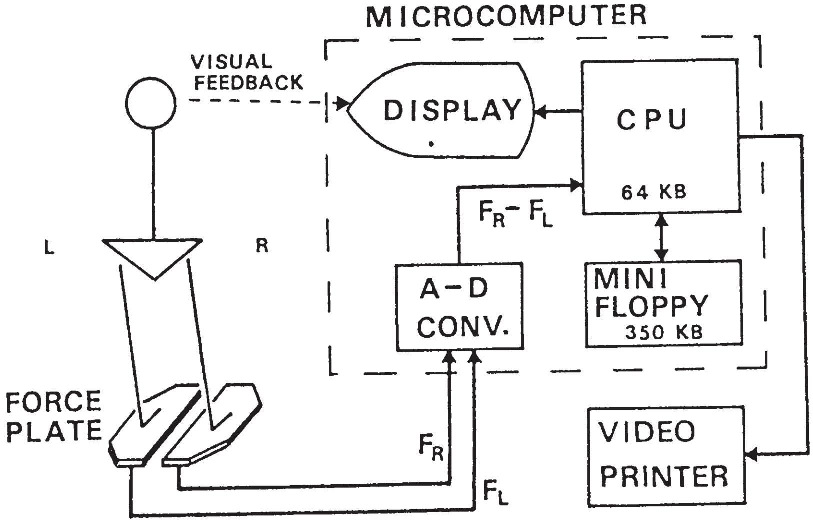


Fig. 1. Diagram of weight balance analyzer

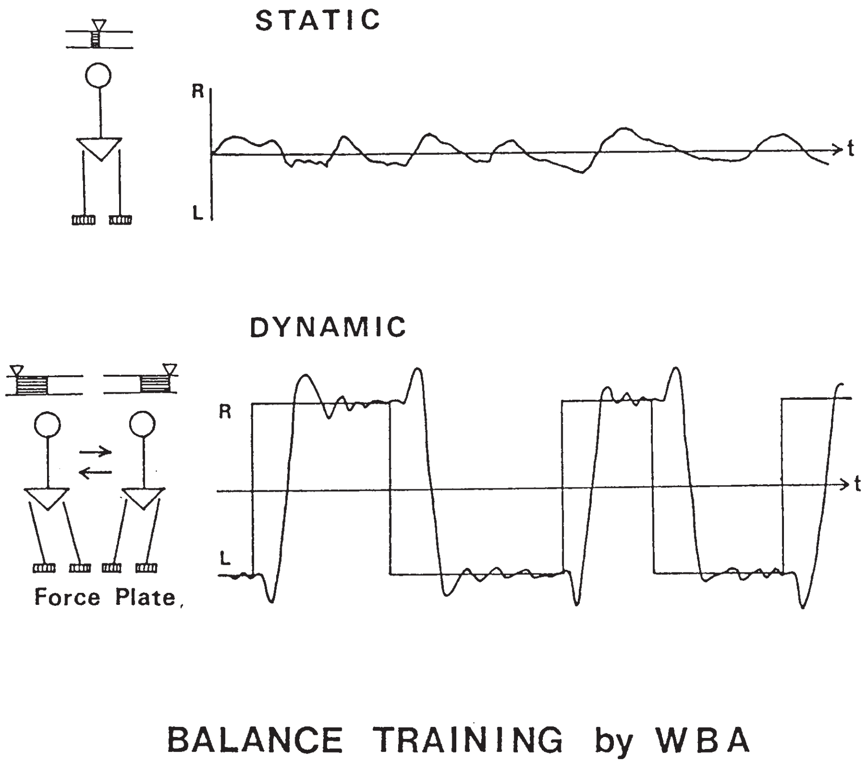


Fig. 2. Balance training by WBA

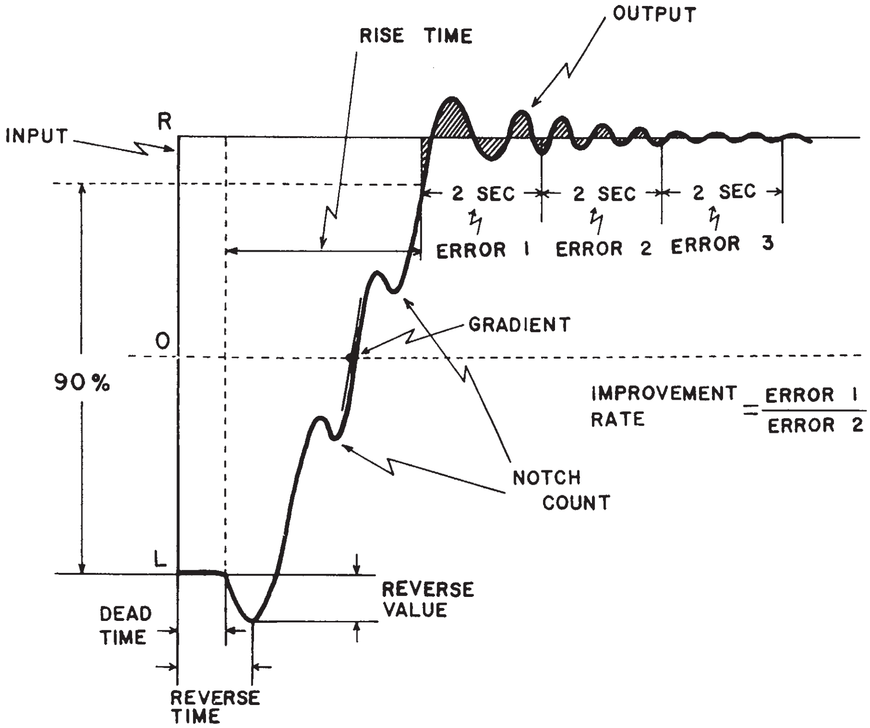


Fig. 3. Parameters of step response

disabled through measurement of tracking characteristics in standing (Fig. 1). This device also could be used as a training machine for every stage of mature gait (Fig. 2). The patients were evaluated and trained for static standing balance as base of walking, right to left and forward to backward body weight shifting according to reference loads which is shown in the display system as a reference signal (dynamic balance). This device was used either in a sitting or a standing position. Moreover, it could be applied from the extremely early stage of rehabilitation, if a chair situated separated in its center was used for the measurement and training. Reference loads was set at 5% intervals from 10% to 95% of body weight and reference signals appear for the right and left sides alternately at random intervals. Examinees made an effort to minimize differences between input given as a reference and output of their own (unilateral body weight shift). Reference signals were shown with square waves and the waves of weight shift of examinees were presented slightly later to follow them. Differences between the two were calculated by microcomputer, and five values of each of ten variables (details in next section), their means and unbiased standard deviations were recorded for each of the right and left legs (Fig. 3).

2. *Measurement in a below-knee amputation*

The results obtained by WBA three weeks after operation (independent standing) were compared with those three months after (independent walk) in a right leg below-the-knee amputee with a below knee prosthetics by Labour Accident Prosthetics and Orthotics Center. The results of body weight shifting from the sound leg (left) to the prosthetic leg (right) three weeks after operation were shown in the upper part of Fig. 4. These were compared with those three months after operation shown in the lower part of this figure and both were observed for

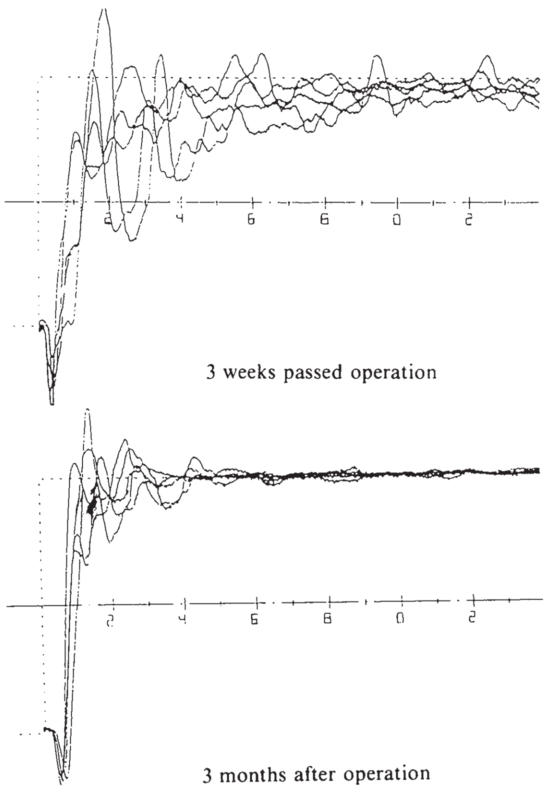


Fig. 4. Postoperative evaluations of right a below-knee amputee amputee by WBA

significant differences by t-test.

a) *Reverse value*: This means the reflected reverse reaction force in the direction opposed to that of the reference signal which was observed immediately after the onset of reaction. It was 27.84 ± 8.17% of body weight in thirty normal subjects. In the amputee it was 9.43 ± 2.87% after three weeks, but 20.18 ± 1.04% after three months, demonstrating an increase in reverse reaction force.

b) *Notch count*: This means the number of transient returns observed during rise reaction (lateral body weight shifting). The smaller the notch count value, the higher is the smoothness of shifting. In the amputee case mentioned earlier, it decreased from 1.4 (3W) to 1.00 (3M), but no significant difference was noted.

c) *Gradient (gradient-force)*: This is the gradient at the midpoint of rise reaction and means the force of body weight shifting. In the present case, it increased significantly from 95.2%/sec. (3W) to 269.0%/sec. (3M), showing that body weight was shifted with force to the prosthetic leg after three months.

d) *Rise time*: This means the time required for lateral body weight shifting from the single stance to another leg. In this case, it decreased from 2.57 seconds (3W) to 1.34 seconds (3M), showing an increase in the quickness of response.

e) *Error (supporting ability of body weight-fluctuation)*: Fluctuation of waves after body weight shifting represents supporting ability and stability of a load-bearing leg. Fluctuation of waves was markedly small in legs with high supporting ability. The sum of the absolute values of tracking errors for the initial 2 seconds after signal attainment was regarded as Error 1. A certain degree of error (6.25% of body weight in normal persons) usually occured because of overshooting. The sum of errors for the subsequent 2 seconds was regarded as Error 2 and that for an additional 2 seconds as Error 3. Error 1 showed no difference between 3W and 3M in the present case, but Error 2 and 3 decreased significantly.

3. *Dynamic standing equilibrium training combined with block-break game (Fig. 5)*
Monotony, scolding and excessive encouragement should be avoided in motion training.

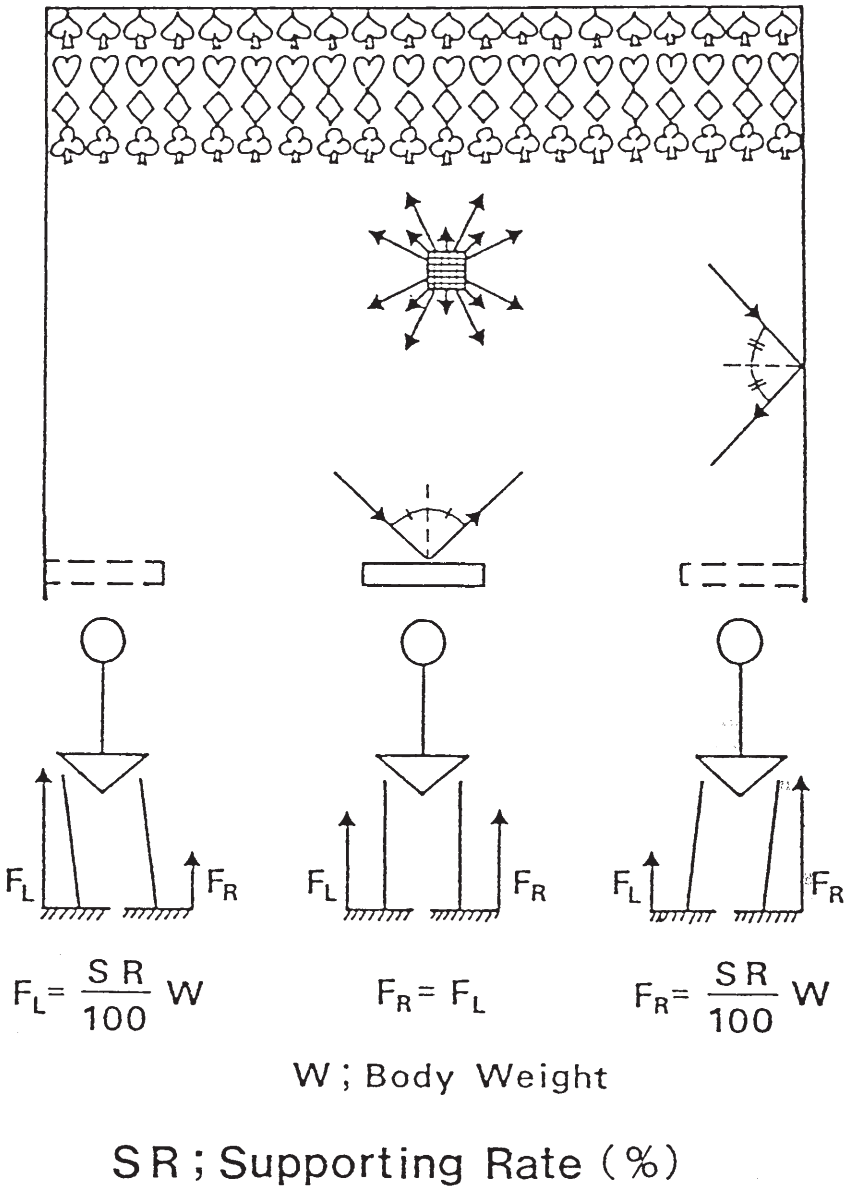


Fig. 5. Block-break game by WAB

Desire for training on the part of the examinee was often enhanced when training was carried out enjoyably in combination with play, sport or game. For this purpose the block-break game was programed and combined with WBA. In this game, the racket on the TV screen moved right and left with the shifting of the examinee's body weight center. The racket remained in the center of the TV screen when the load was supported equally with both legs, but moved to the right when body weight shifted to the right leg. Reference loads was set at 5% intervals from 55% of body weight. Balls were hit with the racket against the wall of blocks and points were scored according to the number of blocks broken.

Using dynamic standing equilibrium training combined with the block-break game, quickness of body weight shifting and ability of subtle regulation of loads were improved enjoyably.

Floor Reaction Force Plate

A large serial force plate (Anima Co., Ltd.) was connected to the micro-computer to measure the serial floor reaction force of several steps. In the description of waves, the time of each step was normalized and the stance phase was standardized to 100%. However, actual measurements in the stance phase were obtained accurately and recorded.

1. *Force components in three-dimensional directions*

Fig. 6 shows three force components in a right hemiparesis patient with mild equinus gait. In this figure, the abscissa represents the stance period of one step normalized to 100% and the ordinate represents the percentage of the patient's body weight. Force components of the left leg are shown in the left column and those of the right leg in the right one. In addition, vertical force (Z) is shown in the upper part, antero-posterior force (X) in the middle part and medio-lateral force (Y) in the lower part. In the right vertical force wave figure, 1, 2 and 3 represent the toe-off period (values of a) and the heel-contact period (values of b) of the left sound leg. Accordingly, the time between values of a and b corresponds to a single stance period. In this figure, the mean stance period of the left sound leg is 1.1 seconds and that of the right affected leg 0.9 seconds. In addition, the mean single stance period of the left sound leg is 0.57 seconds and that of the right affected leg 0.47 seconds, showing that standing supporting time is shorter in the affected leg.

When the way of loading is compared between right and left legs, loading to the affected leg lacks smoothness and 100% loading occurs very late (about 30% time of the stance cycle) as compared with the sound leg (18% time). Moreover, about 50% of body weight is loaded suddenly to the forefoot at 10% time of the stance cycle and whole body weight is then loaded gradually after transient decrease.

After whole body weight is loaded (from 30% of the stance cycle), excessive loading above 100% occurs and extra energy is consumed in addition to loading to the affected leg (muscle weakness). The 100% or more loading period (waves slightly below 100% are not taken into account) becomes short to compensate such overloading. It is (73% − 30%) × 0.9 sec. = 0.39 in the affected leg, while (85% − 18%) × 1.1 sec. = 0.74 sec. in the sound leg. The onset of whole body weight loading occurs late and its duration is short. Accordingly, decrease in loading occurs early (73% time of the stance cycle) in the affected leg as compared with the sound leg (85% time).

In the figure of antero-posterior force (X), (+) represents deceleration force and (−) represents acceleration or driving force. The affected leg is characterized by the acceleration force attaining its peak early (at 73% time of the stance cycle) as compared with the sound leg (at 90% time) and what is not noted after 87% time, showing absence of backward kick.

In the figure of medio-lateral force (Y), (+) represents outer (O) floor reaction force, i.e., inner action force, and (−) represents inner (I) floor reaction force, i.e., outer action force. The

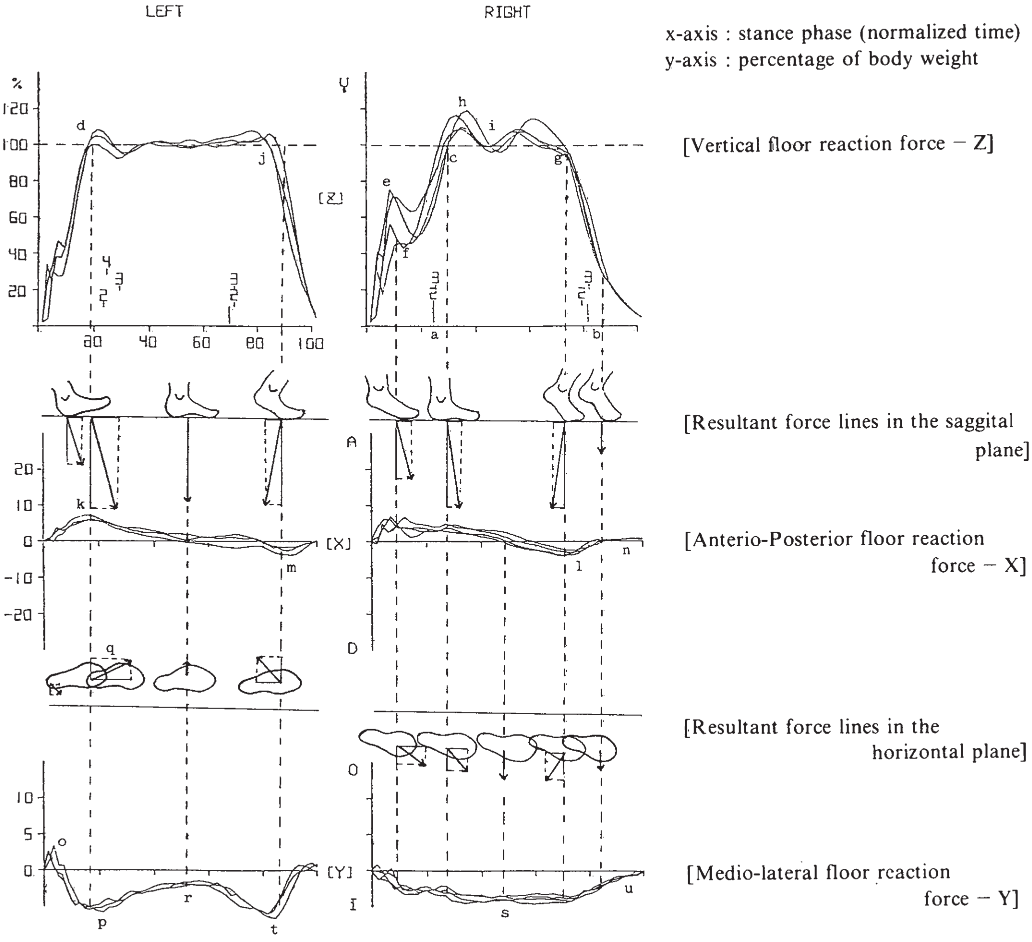


Fig. 6. Floor reaction forces of right hemiplegic patient.

small and sharp inner action force noted at 5% time of the stance cycle in the sound leg means that force acts in the varus position at heel contact. It is replaced by the vulgus position immediately, followed by shifting to foot flat. On the other hand, foot contact occurs from the forefoot in the right affected leg and small waves at heel contact are not observed. The next peak in the sound leg occurs at 18% time of the stance cycle, synchronizing with the peak of deceleration force (X). In this period, foot flat begins and whole body weight is loaded (Z). When the resultant force of X-Y is observed, it is known that the load is added anterolaterally. This peak is not observed in the affected leg. At 50% time of the gait cycle of the sound leg, outer action force becomes small, suggesting that the center of body weight passes near the contact plane of the left leg. In the affected leg, outer action force becomes maximum, assuming a pattern which is directly opposed to that of the sound leg. The peak of outer action force at 85% time of the stance cycle in the sound leg reflects outer kick before toe off, and body weight seems to be shifted to the right leg by this kick. Since this peak of the outer kick is absent in the affected leg, it is found that body weight is shifted slowly from the affected leg to the sound one.

2. *Three-Dimensional Goniometer for the Hip Joint*

Since the hip joint performance flexion-extension, adduction-abduction, internal-external rotation and circumduction as complex motion, it may be said that this joint fundamentally performs three-dimensional motion. Angle variations during walking measured by a three-dimensional goniometer (devised by the authors) are processed and described by micro-computer (Photo. 1).

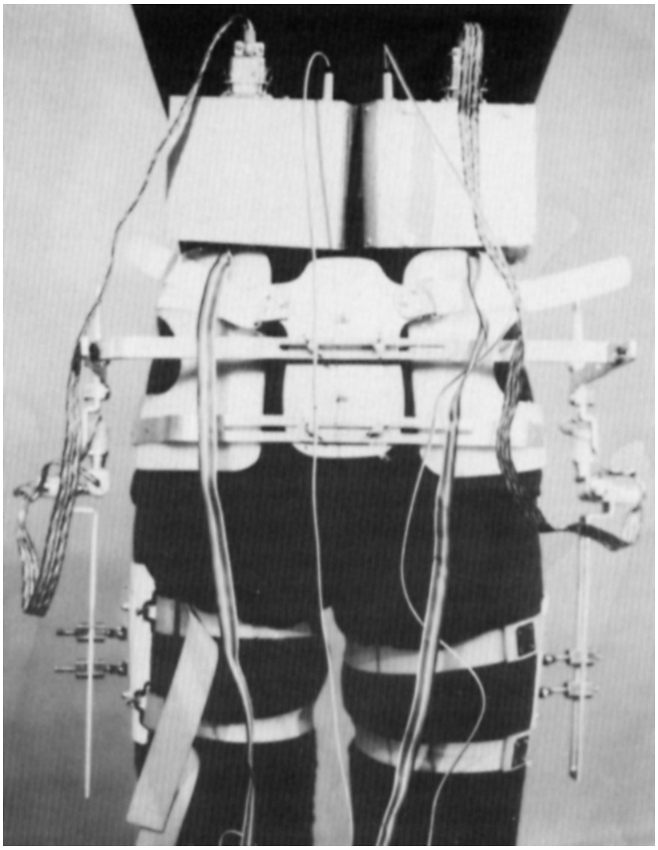
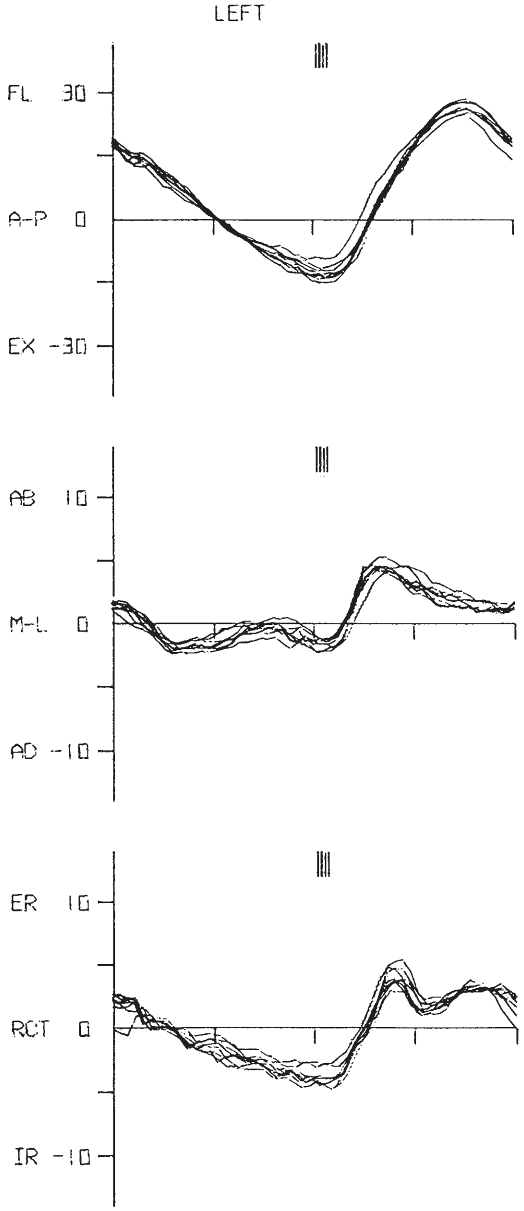


Photo. 1 Three-dimensional goniometer



||||| : opposite heel contact
gait cycle time : 1.07 ± 0.02 sec.
maximum flexion
angle: 29.6 ± 1.0°
time : 86.5 ± 1.7%
maximum extension
angle: 10.8 ± 1.5°
time : 50.1 ± 1.4%

maximum abduction
angle: 6.4 ± 0.5°
time : 63.4 ± 2.4%
maximum adduction
angle: 3.0 ± 0.5°
time : 20.2 ± 4.0%

maximum external rotation
angle: 3.1 ± 0.5°
time : 66.7 ± 1.3%
maximum internal rotation
angle: 2.8 ± 0.2°
time : 54.3 ± 15.0%

Fig. 7. Angular changes of the normal hip joint by three-dimensional goniometer

Fig. 7 shows three-dimensional angles of the hip joint in a normal person. In this figure, the time between heel contact of both legs is normalized to 100%. The walking cycle of this case is 1.07 ± 0.02 sec. and it includes 8 steps. The hip joint is in the flexion position of about 20 degrees at heel contact, but this angle decreases immediately and the maximum extension of 10 degrees is attained soon after the completion of contralateral heel contact (50% time). During this stance phase, the hip joint continues to be in the adducted position and the peak of the maximum adduction of 3.0 degrees is attained when whole body weight is loaded immediately after foot flat (about 20% time). Waves of adduction-abduction angles during the stance phase (0–60% time) are similar to adduction-abduction force waves of previously described floor reaction force (Fig. 6) or action force in the sound leg. It may be quite reasonable that medio-lateral body weight shifting and adduction-abduction angles of the hip joint assume the same variation pattern. In internal-external rotation, the internal rotation angle of the hip joint increases gradually throughout the stance phase and attains the maximum angle of 2.8 degrees at around heel off (54.3% time), which is almost synchronized with the m and t waves of previously described floor reaction force (Fig. 6).

After contralateral heel contact, the stance phase is shifted to the swing phase through the double stance phase for about 10%. In the swing phase, which begins at about 60% time, flexion of the hip joint increases greatly, attaining the maximum angle of about 30 degrees at 87% time. It decreases to 20 degrees before heel contact. In adduction-abduction, the abducted position (the maximum abducted angle of 6.4 degrees at 63.4% time) is taken in the swing phase and the external rotation position (the maximum external rotation angle of 3.1 degrees at 66.7% time) is maintained throughout the swing phase. In other words, extension, adduction and internal rotation are combined in the stance phase, while flexion, abduction and external rotation are combined in the swing phase. When angles of the knee and ankle joints are taken into account, the extensors of the knee joint operate in the stance phase, resulting in high incidence of extension of the knee joint, and the flexors of the ankle joint operate only in the stance phase. In the swing phase, however, flexion of the knee joint attains its peak and the dorsal flexors of the ankle joint also operate well in this phase. These facts suggest that normal walking also shows a tendency of extension synergy in the stance phase and that of flexion synergy in the swing phase. In the presence of central nerve disturbance, these synergic tendencies are enhanced markedly and scissor's gait (hip adduction gait) in the stance phase and circumduction gait (hip abduction-external rotation gait) in the swing phase are often experienced clinically. The three-dimensional goniometer for the hip joint seems suitable to grasp these walking abnormalities both graphically and quantitatively (Table 1).

Table 1. Angular changes and muscle actions of lower limb during walking

Walking / Joint	Stance phase (extensor synergy pattern)	Swing phase (flexor synergy pattern)
Hip joint	Extension -adduction -internal rotation	Flexion -abduction -external rotation
Knee joint	Action of quadriceps, much extended position	Maximum flexion peak
Ankle joint	Action of plantal flexors	Action of dorsal flexors

DISCUSSION

Micro-computers are in widespread use even among families and children and they should also be utilized now positively in the field of rehabilitation. Since sufficient information can not be obtained by a single method, it is important to combine various techniques to collect sufficient data. In addition to the devices described above, measurement of sole-contact order by foot switch, and serial measurement of separated vertical force from the forefoot and rearfoot by foot sensor are being carried out at the Nagoya University Hospital. It is important that accumulation, utilization and analysis of these data can be accomplished conveniently and within a short time. It is also important that the data printed out are graphic and can be understood easily by intuition. The weight balance analyzer devised by the authors can be used for the measurement of dynamic standing balance ability as well as for training from its initial stage. Therefore, this device is advantageous because it can be used not only for analysis but also for training. Graphic and quantitative evaluation of the process of training encourages patients and, moreover, provides examiners with objective data. By introduction of the block break game, monotony is avoided and training can be performed enjoyably, competitively and positively. The force plate for the measurement of floor reaction force is a representative device for kinetic motion analysis during walking. The menu of our measuring program of floor reaction force includes 8 items and is characterized by the fact that examiners can choose the necessary one from among them. The 8 items included are as follows: (1) serial original waves of floor reaction force, (2) superimposed waves of floor reaction force, (3) waves of loading velocity (differential waves of each floor reaction force), (4) paths of center of pressure, (5) action force lines, (6) Lissajou's wave types, (7) COP velocity and acceleration of center of pressure, and (8) paths of center of pressure for each of the right and left legs. These can be selected easily with a switch of the micro-computer even by physical therapists. By connecting a goniometr to the same micro-computer, changes of three-dimensional angles of the hip or knee joint during walking can be grasped. In contrast to the floor reaction plate for kinetic analysis, this three-dimensional goniometer is a representative device for kinematic analysis. The data obtained by these three representative devices are processed by the on-line system of one micro-computer to provide objective evaluation for motion analysis. These data also serve as guidelines of training for both examiners and examinees, providing the former with scientific justification of training and the latter with encouragement.

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