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An investigation of kinematic and kinetic variables for the description of prosthetic gait using the ENOCH system

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Abstract

Gait patterns, joint angles, floor reaction forces and joint moments during walking were investigated for normal subjects and above-knee and below-knee amputees.

The investigation showed that the hip-knee angle diagram as well as different symmetry diagrams (e.g. left knee angle versus right knee angle) provide an easily interpreted means of evaluating abnormalities in the gait pattern. It was further concluded that a combined gait pattern-force vector diagram is valuable for the evaluation of the joint moments.

Floor reaction forces and muscular moments at the joints were also included in the analysis. The joint moments at the knee were quite different for both above-knee and below-knee amputees as compared to the normal subjects. Some interesting trends were also found concerning the knee stability of the amputees.

A system called ENOCH was used for the measurement and analysis. This system consists of a minicomputer connected on-line to equipment for measurement of displacement (Selspot) and floor reaction forces (Kistler). A graphic computer terminal (Tektronix) was used for the result presentation.

Introduction

It is generally agreed that there is a need for quantitative analysis of human gait for the evaluation of abnormalities in the locomotor apparatus. Many different types of equipment have been developed over the years. Among these are:

Goniometer systems which give joint angles that can be used to characterize the gait pattern. Here the data are given in direct electrical form.

Cinematography which gives a kinematic description of the gait. In this case quantitative evaluation is very time consuming and expensive.

Force plates which give the floor reaction forces.

TV-based systems which give the position of selected landmarks on the body. With the cameras connected on-line to a computer it is possible to make quantitative analyses in direct connection to a measurement. However, current systems have limited resolution and sampling rates.

Systems based on position sensitive photodetectors. This kind of equipment makes it possible to obtain cartesian coordinates for selected landmarks with a precision and sampling rate that is superior to the TV-based systems. A drawback is that light sources must be carried by the subject.

Reports on practical usage of all types of equipment mentioned above are extensively found in the literature, except for the last type which can hardly be found at all. It is this kind of equipment that was used for the present investigation.

In conjunction with such measurements, analyses of the data based on mathematical models of the human body (McGhee, 1981; Oberg, 1974) are used in research laboratories but can hardly be found in clinical use.

Method and material

A minicomputer based system—called ENOCH—was used for the measurements and analyses (Gustafsson and Lanshammar, 1977). In this system (Fig. 1), an optoelectronic device, Selspot, with position sensitive photodetectors is used for kinematic data collection. Ground reaction data are obtained from a Kistler force plate. Output of result diagrams are made on a graphical computer terminal with a hardcopy unit or in tabular form on a line printer.

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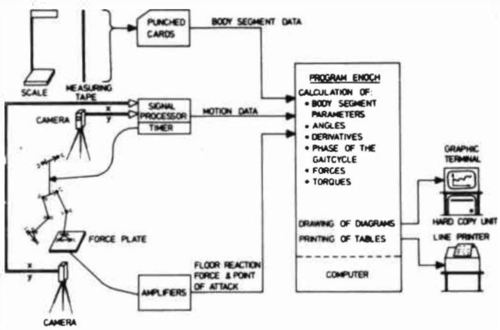

Fig. 1. Schematic diagram of the ENOCH gait measurement and analysis system.

Two Selspot cameras were used to obtain kinematic data for both legs. Landmarks (light emitting diodes) were placed on the shoulder, hip joint, knee joint, ankle joint, heel and toe base for both sides (Fig. 2). The measurement area was approximately 3 m × 3 m allowing for the registration of 3 steps in each measurement. Data was collected at the rate of 158 Hz. The standard deviation of the measurement noise was 2 mm and the systematic coordinate error was estimated to be less than 2 cm.

The displacements of the centre of mass for the different body segments in the model, HAT (that is head, arm, trunk), thighs and shanks, were calculated from the measured coordinate data. The required body segment parameters were obtained according to a method based on data from Drillis and Contini (1966) and Chandler et el (1975). Absolute angles for the body segments and relative angles at the joints were obtained from the linear displacement by straightforward application of trigonometric relations.

The velocities and accelerations of the different body segments were calculated by numerical differentiation of the displacement data. The differentiation procedure is described in Gustafsson and Lanshammar (1977). It is a design based on minimization of the total error in estimated derivatives where a systematic error component is obtained from the rest term in a Taylor series expansion of the signal, and a stochastic error component results from uncorrelated measurement noise added to the signal. The structure of the algorithm is a linear finite impulse response (FIR) filter where the filter coefficients are determined by the minimization mentioned above.

Gait phase changes were automatically determined by application of algorithmic tests on the velocities of the feet landmarks.

Finally, forces and moments at the joints were calculated by application of the Newtonian equations of motion.

It should be noted that only planar motion was included in the analysis. Further the shank and foot was treated as one rigid body.

Measurements were made on 5 male persons with below-knee (BK) and 3 male persons with above-knee (AK) amputations. For reference, measurements were also made on 4 normal subjects.

Results

Many different types of diagrams were studied with respect to their ability to characterize the gait.

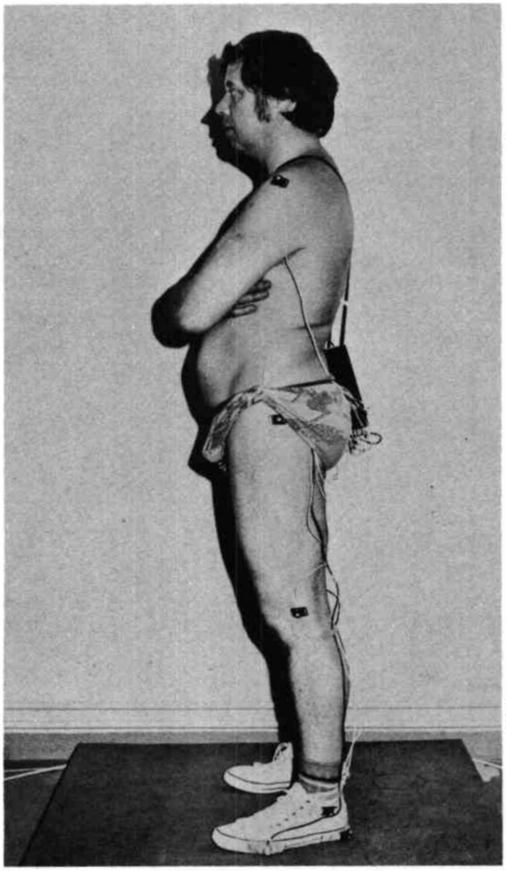

Fig. 2. Light emitting diodes (landmarks) mounted on an AK patient.

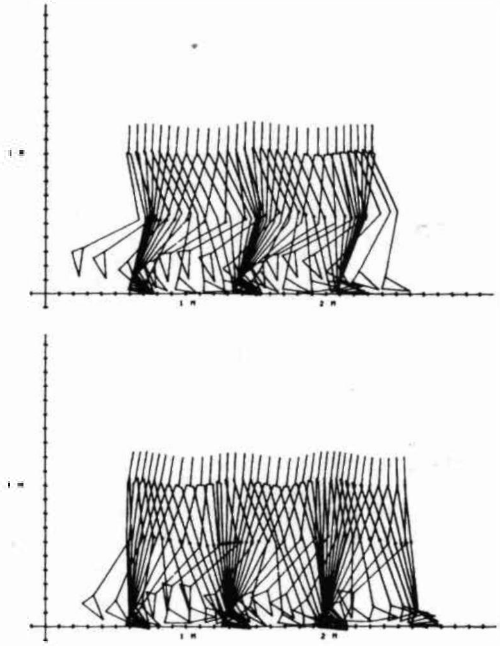

Fig. 3. Stick diagrams showing the geometrical gait pattern. Top, normal subject, bottom, right leg above knee amputee.

Figure 3, top, gives a description of the geometrical gait pattern. The force plate was located between 1·2 m and 1·8 m on the horizontal axis. The right leg is marked with a small square on the right knee. In this case the subject was a normal.

Figure 3, bottom, shows the same diagram for a right leg AK amputee, wearing a prosthesis with Blatchford stabilized knee and a Greissinger foot. The asymetry in the gait is obvious.

The well known hip-knee angle diagram (Lamoreux, 1978) is an already rather well established joint angle description. For an AK amputee Figure 4, bottom, shows a typical example. As can be seen the knee angle is zero during the stance phase, and the diagram looks very much like a triangle. This is in sharp contrast to the corresponding diagram for normal gait (Fig. 4, top), where the knee is flexing also during stance.

Figure 5, the knee-knee angle diagram, provides a means to evaluate the gait symmetry between the left and right side. If the gait is symmetric, which is the case in Figure 5, top, the curve is symmetric about a line with slope 1. For AK prosthetic gait the curve is not at all symmetric, which can be seen in Figure 5, bottom.

Another type of diagram that was studied were plots of joint moment versus time. In Figure 6, top, the knee moment for a normal subject is plotted. In the diagram a positive moment means a flexing muscle moment. During stance phase the moment is alternating between a flexion and an extension moment. For the prosthetic side of the AK patients, the corresponding diagram looked like that in Figure 6, bottom. In this case there is a flexion moment during the entire stance phase. This moment is due to the extension stop in the knee mechanism.

This observation can be understood by looking at Figure 7, where the floor reaction force is plotted on a stick diagram of the gait pattern for 5 points of time during stance phase. This diagram output was specially designed for this investigation.

Inertial forces have very little influence on the joint moments during stance. Therefore the joint

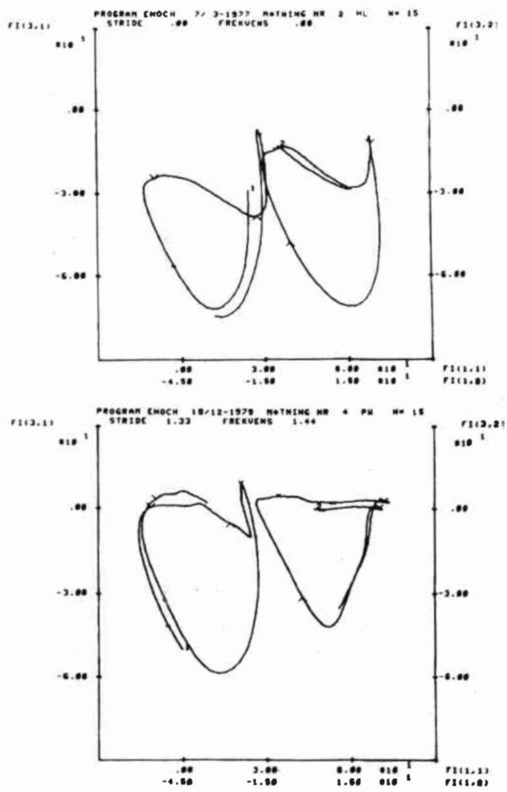

Fig. 4. Hip-knee angle diagram for left and right leg. Top, normal subject, bottom, right leg above knee amputee.

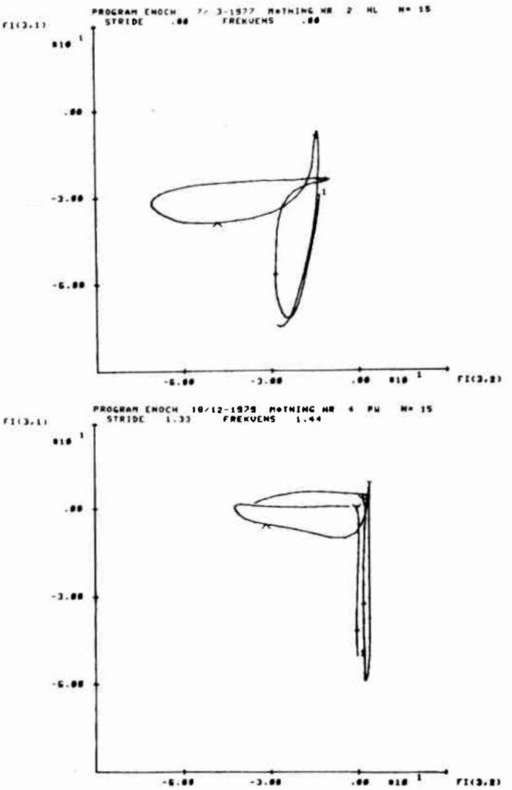

Fig. 5. Knee-knee angle diagram. Top, normal subject, bottom, right leg above knee amputee.

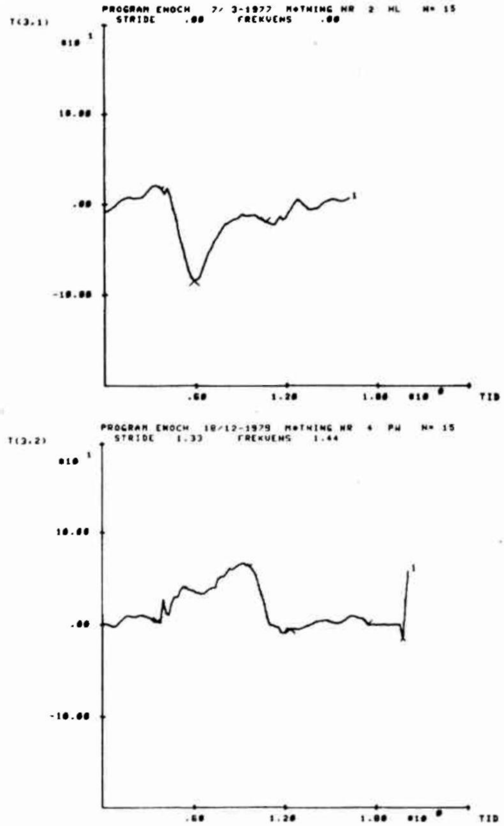

Fig. 6. Knee moment versus time. Top, normal subject, bottom, right leg above knee amputee.

moment resulting from the floor reaction force must be balanced by counteracting muscular moments.

In Figure 7, top, the direction of the floor reaction force is such that it results in a flexion moment in the middle of stance phase. Therefore the muscular moment at the knee joint is changing sign and exhibits an extension moment during most of the stance phase. This is in accordance with the muscular moments shown in Figure 6, top.

In Figure 7, bottom, it can be seen that the floor reaction force gives an extension moment during the entire stance phase (except at toe off). This explains the flexion moment observed in Figure 6, bottom.

The knee joint of this investigated AK amputee was fitted with a weight bearing controlled knee joint (Blatchford). However, since the person was stabilizing the knee joint by contracting the hip an extension moment at the prosthetic knee joint was produced. Therefore the knee lock was not used at all by this subject during the experiments.

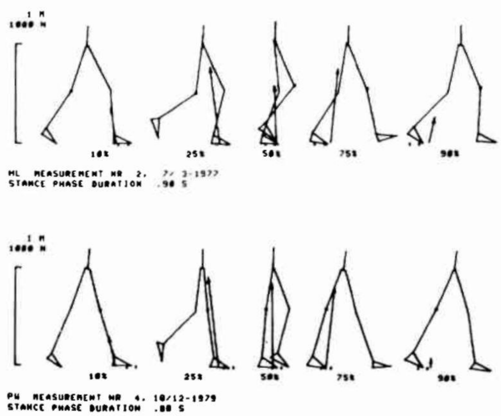

Fig. 7. Gait pattern-force vector diagram. Top, normal subject, bottom, right leg above knee amputee.

Conclusions

This investigation has demonstrated that simultaneous measurements of the displacements of body markers and floor reaction forces combined with on-line computer analysis is a powerful tool for the assessment of gait dynamics. By the utilization of positive sensitive photodetector based equipment, the displacements of the body markers can be determined with high precision and at relatively high sampling rates. Further by using computer graphics, the presentation of results can be made easy to interpret as, for instance, in the combined gait pattern-force vector diagram.

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