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TECHNICAL NOTE

A PROCEDURE TO VALIDATE THREE-DIMENSIONAL MOTION ASSESSMENT SYSTEMS

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Abstract—The automation provided by computer-assisted motion-tracking systems allows for three-dimensional motion and force analysis. These systems combined with mathematical modelling are able to analyse quickly the intricate dynamics of human movement. Understanding the limitations of human motion analysis as performed by the present measurement techniques is essential for proper application of the results. It is necessary to validate the analysis system prior to subject testing. This paper provides a validation of an optoelectric motion-tracking system used in a dynamic knee assessment study. While the validation is shown with one particular system only, it is suggested that all systems used in two- or three-dimensional motion analysis should be tested similarly in the actual configuration used. Three simple mechanical representations of the human knee have been used in this validation. The first model provided an understanding of the source and behaviour of the error introduced to the accuracy of defining a vector between the recorded coordinates of two markers. The other two models investigated the effect of processing methods specific to the knee analysis project. Separating the markers by at least 180 mm is recommended to produce stable vectors. Relative joint angles could be calculated in all three planes of rotation. The error in calculating flexion and longitudinal rotation was less than 2.0°, while calculating adduction introduced errors of 4.0°. Force calculations were found to be within 8%. The system behaviour was found to be consistent within the calibrated volume about the force platform. Simple mechanical models combined with straightforward procedures can provide validation in terms of clinically relevant parameters.

INTRODUCTION

Although manufacturers of motion analysis systems provide data about the resolution of these systems, it is often difficult to interpret exactly what these data mean in an experimental or clinical context. One of the first attempts in the calibration and verification of an optoelectric system was performed by Mann and Antonsson (1983) and Antonsson and Mann (1989). This study examined translation, rotation and kinetic measurements. The measured kinematic data were within 1 mm. The noise and all errors with the kinematic data contributed less than 5% to the resulting forces as compared to a theoretical two-degrees-of-freedom pendulum. Samuelson *et al.* (1987) also studied the accuracy of an optoelectric system with the use of mechanical models. The dynamic test was mostly qualitative in nature, relying on the well-defined motion of the model. Scholz (1989) also recognized that a careful examination of information about the resolution, reliability and validity of one's chosen motion-tracking system is crucial. This prompted a quantitative and functionally relevant evaluation of the reliability of an optoelectric system for measuring angular motion. Static angles were calculated to within 0.5°; however, the error increased as the goniometer was rotated to 45° away from the cameras. This study still left unanswered questions regarding the accuracy beyond a one-degree-of-freedom system and thus did not exploit the three-dimensional nature of the optoelectric system. Also, the study provided no assessment of kinetic analysis.

The validation of a gait analysis system must be done at the data entry level as well as at the data processing level. It must include both a kinematic and a kinetic assessment. The results should be easy to interpret in terms of measured parameters used in a clinical context.

The goals of this study were to develop simple mechanical models to test different aspects of any type of motion-tracking system. The procedures and analysis techniques were, however, applied only to one particular system used for one specific study.

METHODS

Mechanical models representing various elements of knee analysis can provide known and exact behaviour.

Model 1

The first of the three models was designed to assess the optoelectric system, independent of the data processing methods used for knee motion analysis. The goal was to evaluate the system at the level of the three-dimensional coordinate identification. The input data consists of the three-dimensional coordinates of each of the markers or LEDs that have been placed on the body of interest. To describe the three-dimensional motion of a body in space requires identifying an array of at least three non-collinear markers fixed to the body. For our study of the knee, one array is used for each of the thigh and shank segments. Each array consists of two LEDs on the lateral side of each segment, and a third on a rigid extension or probe attached to each segment. The primary function of these markers is to establish localized coordinate systems fixed to each segment. This is accomplished by defining vectors with the LEDs as endpoints. The model was used to assess the ability of the optoelectric system to locate a vector in space. The objective was to examine LED separation distance as it is affected by the presentation of the LEDs to the cameras. The shortest

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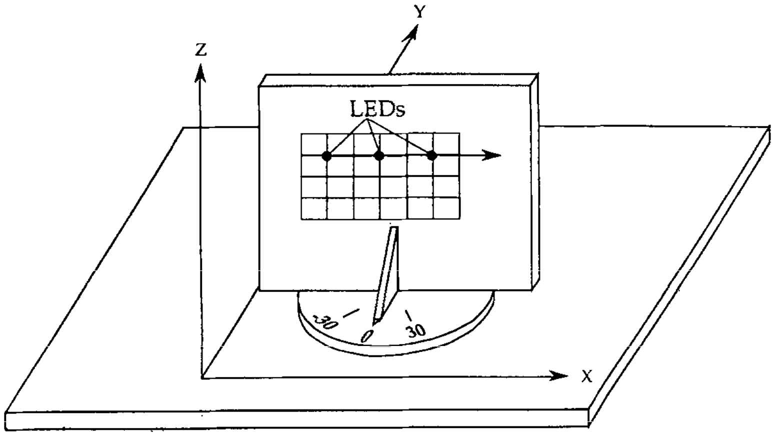

Fig. 1. Model 1 consists of a plexiglass plate mounted on a rotatable base. A 20 mm grid was scribed onto the plate allowing placement accuracy of the LEDs to within 1.0 mm. The base was scribed at five increments and instrumented with a potentiometer to record the angle at which the LEDs were presented to the cameras. Calibration of the potentiometers provided angular measurements to within 0.5.

distance with minimum error over the widest range of motion would be chosen as the length of the probe. The model consisted of a plexiglass plate mounted on a rotatable base (Fig. 1). Ten LEDs were placed collinearly, so that combining one LED with each of the others produced nine estimates of a single vector using separation distances of 20 to 220 mm. The location of this vector was defined by the angle between it and the global X-axis. Five angles of presentation within the range of $\pm 30°$ were chosen and, for each run, one of the five positions was chosen at random using the random number generator of a Hewlett-Packard 15C calculator. Each position was then sampled at 50 Hz for 1 s. Two replications of each static position were performed.

Model 2

The second model was used to evaluate the calculation of three-dimensional relative joint angles. To accomplish this, a mechanical representation of the human leg was designed and built. The model consists of two segments representing the tibia and the femur joined by a mechanical knee (Fig. 2). The 'knee' is a three-degree-of-freedom joint simulating the three rotations of the human knee. By securing the rotation axes the model could be held in one position, or moved with one, two or three degrees of freedom, allowing comparisons between static and dynamic behaviour.

There is a wide variety of methods available to describe the configuration and motion of a joint. The differences that exist between them make some more suitable than others in certain circumstances (Andrews, 1984). The floating-axis method, proposed by Grood and Suntay (1978) and Suntay and Grood (1983), is a technique used to describe the relative spatial position of two rigid bodies which constitute a joint. The ease of calculation and the clinical relevance of the orientation angles it produces make it quite appealing.

Model 3

One important aspect of the human motion study is the assessment of forces at the knee joint, specifically, to determine the dynamic mechanical parameters in the knee joint during gait and stair-climbing. This requires a three-dimensional dynamic algorithm which incorporates information from gait analysis, force platform, muscle activity and X-rays to calculate the bone on bone forces at the proximal end of the tibia. The algorithm involves double differentiation of the displacement data collected by the system as well as simultaneous solving of the six equations of motion (Li *et al.*, 1992). Unlike the first model, which assessed the system independent of computer algorithms, this model evaluates both the optoelectric system and a complex mathematical model. The measurement of inertial forces during the swing phase of gait relies on approximations of the mass properties combined with accurate motion analysis. The objective of this phase of the study was to investigate the accuracy of inertial-force calculations to identify significant contributors of error and conditions which affect these calculations. To determine the accuracy of the system in measuring forces at the knee, a mechanical representation of the knee was used. The representation was kept as simple as possible in order to yield meaningful and easily interpretable results (Fig. 3). This simulates the motion of the shank about the knee during the swing phase of gait. The dimensions and mass moment of inertia were chosen to simulate the shank and foot of a 70 kg person (Winter, 1979). A strain gauge load cell was built, to measure directly the axial forces and was incorporated into the pendulum close to the rotation axis and calibrated on a load testing machine.

The model was positioned at distances of ± 20 cm from the centre of the 50 cm wide force platform. The angle of presentation to the cameras was also varied between $\pm 15°$ at each of these distances. Forces produced in the pendulum are related to the initial angle it is released from. This angle was varied from 40 to 60° to cover the range of forces expected in the human motion trials (Li *et al.*, 1992). Before velocities and accelerations are calculated, the displacement data are smoothed using a Butterworth filter. It was decided to vary the cutoff frequency from 4 to 8 Hz to assess the sensitivity of the system to filtering. These factors were combined in a fractional factorial experiment involving 26 trials. Factorial designs are widely used in experiments involving several factors where it is necessary to study the joint effect of these factors on a response (Montgomery, 1984). Fractional factorials provide a way to reduce the number of runs required by ignoring high orders of interaction among the factors. In this case, we ignore a four-factor interaction. This allowed estimation of the effect of each factor as well as interactions between the factors. The response for each trial was defined as the average difference between the load cell forces and the calculated forces during the second complete cycle of the pendulum.

RESULTS

Model 1

The orientations of vectors of nine LED separation distances were compared with the true orientation at each of the plate positions. Figure 4 shows the deviation from the true

Fig. 2. Model 2 consists of two rigid segments joined by a three degree of joint. The three rotations simulate those of flexion/extension, ab/adduction and longitudinal rotation as they occur in the human knee. Each axis is instrumented with a potentiometer to record the position of the 'leg' to within 0.5. Each rotation can be secured anywhere within its range of motion. The model is shown with the LEDs attached as they would be in the human subject test.

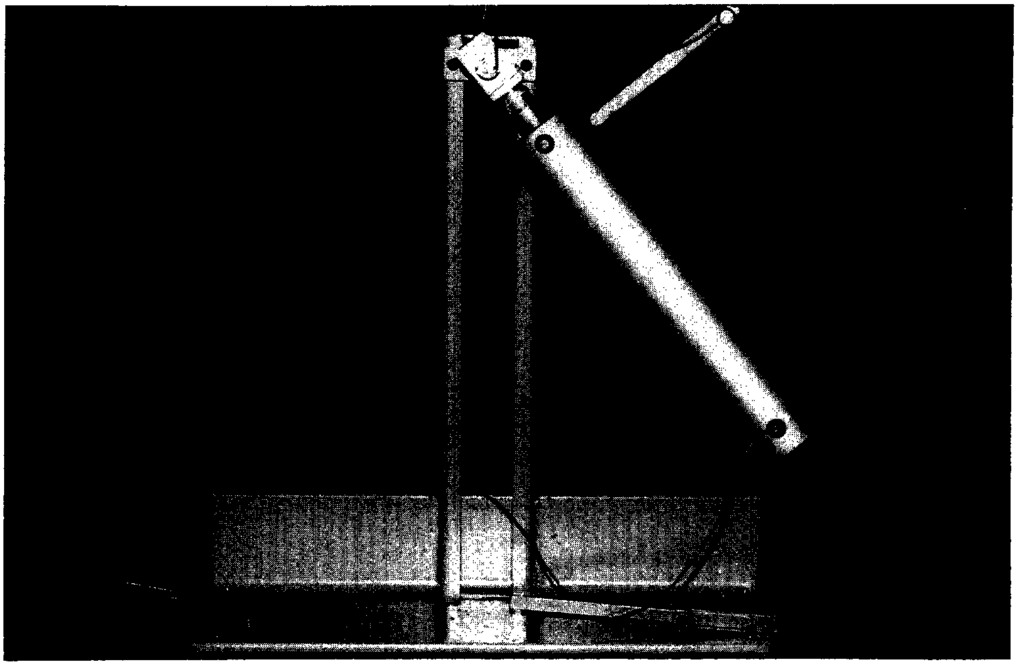

Fig. 3. Model 3 consists of a freely moving compound pendulum with one degree of freedom. The pendulum was instrumented with a strain gauge transducer calibrated to yield axial load. The model is shown with the LEDs attached as they would be in the human subject test.

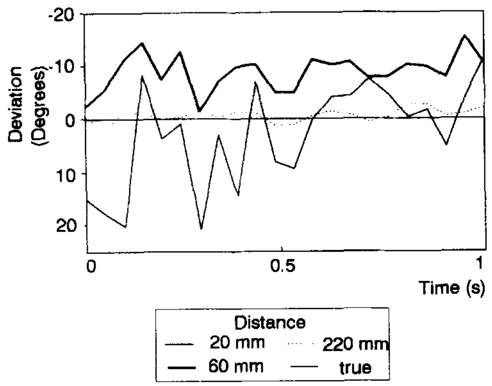

Fig. 4. The error in locating the vector using three separation distances is for one run. Both the magnitude and the variation in the error decrease with the LED separation distance.

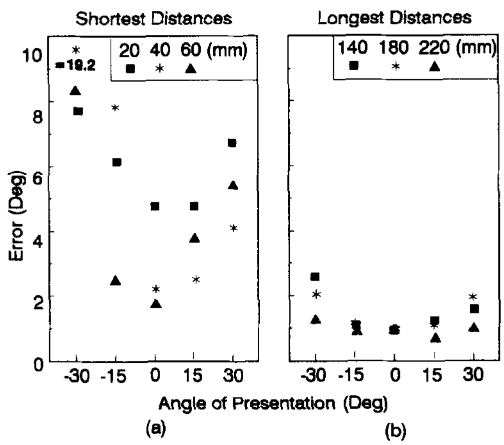

Fig. 5. The error in locating the vector using both the shortest (a) and the longest (b) distances is shown for each plate position.

position for three separation distances for each frame of one trial. Although only one position is shown here, the behaviour is typical of other positions. The shorter distances correspond to a large error magnitude as well as a large amount of variation. The longest distance is close to the true position and variation is minimal. The average error magnitude and the standard deviation of the estimated location were calculated.

To select the optimum distance, analysis of variance techniques were used. Although the positioning of the plate was done in a random fashion, the response associated with all LED separation distances was determined during the same run or plate positioning. Since we were unable to randomize completely both plate positioning and LED separation distances, the data must not be analysed as a factorial. These restrictions on randomization were incorporated into the analysis of variance. Although some precision is lost in comparing positions, this design results in greater precision in comparing LED separation distances than a factorial experiment (Montgomery, 1984). As LED separation distance was the effect of greatest interest, this design seemed appropriate.

Using camera–subject distances of almost 5 m, as required for the human subject study, prevents the use of small LED separation distances (60 mm or less). Figure 5(a) indicates the

Table 1. Optimum separation distance

Separation 180 mm	Error average	Standard deviation
Viewing $\pm 30°$	$< 2.5°$	$< 2.0°$
Range $\pm 15°$	$< 1.5°$	$< 1.5°$

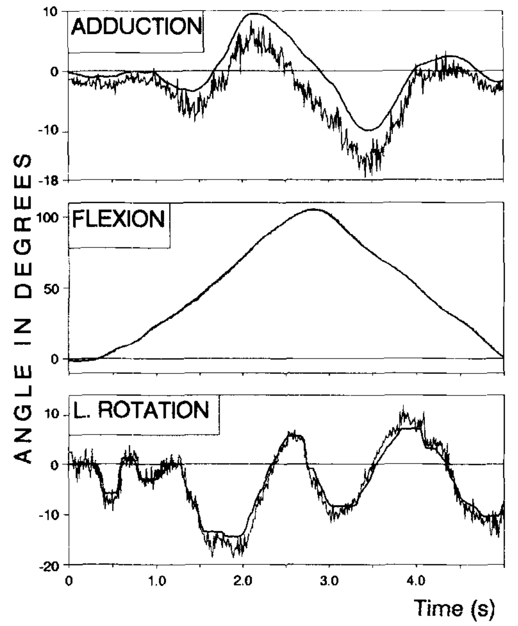

Fig. 6. The relative angles are shown for one of the manual manipulations of model 2. The 'smooth' curves correspond to the potentiometer recordings, while the jagged curves are the calculated angles. These curves were produced by raw data before application of any smoothing techniques.

large error magnitude produced when using these distances. The dependence of the shorter distances on the angle of presentation (position) is shown clearly by the curvature in the data. Figure 5(b) shows how increasing the separation of LEDs allows larger viewing ranges.

The results must be considered jointly in the selection of the optimum separation distance. Confidence intervals placed on the observations of Fig. 4 indicate no significant improvement beyond 180 mm. The viewing range, however, does play a role. More stable results are possible with a slightly decreased viewing angle (Table 1). Vectors can be identified within $\pm 1.5°$ if the LEDs are presented within $\pm 15°$ to the walking direction on the walkway.

Model 2

The floating-axis method was applied to the mechanical knee and compared with the potentiometer measurements. The accuracy of the potentiometers was, after careful calibration, within 1°. The linkage of the knee model was constructed accurately, but it could still have contributed up to 1° of the total errors. No further tests on the knee model or potentiometers were performed, as those errors were well within the desired precision. The simultaneous recordings of the relative rotations of the 'knee' can be seen in Fig. 6. Comparing these results with the static ones reveals similar behaviour for each type of motion analysed. Any differences seen in the error estimates are small enough to be beyond

Table 2. Relative angle calculations (error in degrees)

		Adduction	Flexion	Longitudinal rotation
Static trial	Average	3.8	0.9	1.5
	Standard	2.2	0.5	1.1
Free-motion trial	Average	3.8	0.9	1.5
	Standard	2.2	0.5	1.1

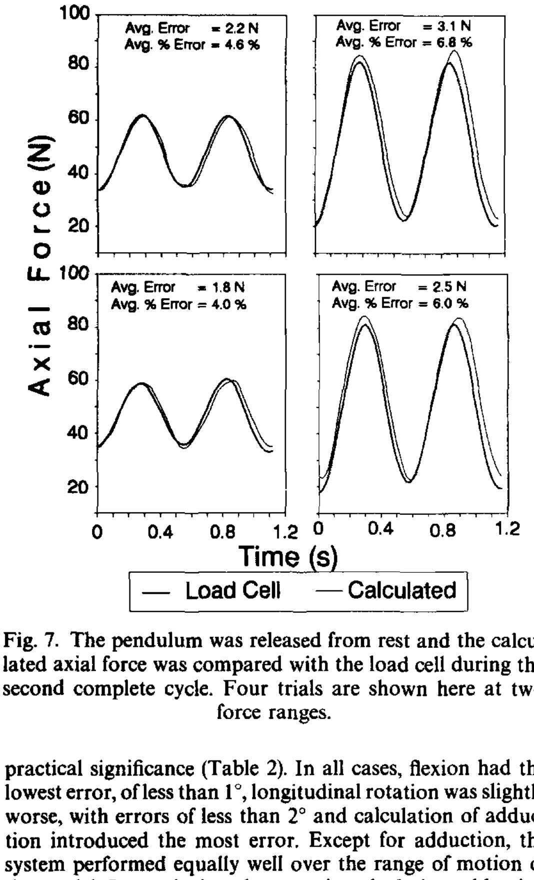

Fig. 7. The pendulum was released from rest and the calculated axial force was compared with the load cell during the second complete cycle. Four trials are shown here at two force ranges.

practical significance (Table 2). In all cases, flexion had the lowest error, of less than 1°, longitudinal rotation was slightly worse, with errors of less than 2° and calculation of adduction introduced the most error. Except for adduction, the system performed equally well over the range of motion of the model. In particular, the error in calculating adduction showed a dependence on the degree of flexion. This may be due to system set-up or possibly an alignment problem with the model but, as yet, we are uncertain. Thus, the system was judged capable of measuring the motions of longitudinal rotation and adduction, with the error bounds shown in Table 2.

Model 3

The data from four of the trials are shown in Fig. 7. The calculated forces agree quite well with those measured by the load cell. The average error over all the factor variations was 3.0 N or 6.7%. The error varied with the initial height of the pendulum, the filter cutoff frequency and the angle of presentation of the model to the cameras. The magnitude of these effects was below that of practical significance and the system behaviour was deemed to be homogeneous. Confidence intervals were calculated for the errors at each of the factor settings. Throughout all factor variations, the axial force could be calculated to within 4.2 N or 8% of the measured force with 95% confidence. It was found that movement towards the front or back of the walkway did not affect the accuracy.

DISCUSSION

It is important, for any type of motion analysis, to know the accuracy in order to interpret the results. To simply indicate the system used and add the manufacturer's specifications is not sufficient. This paper deals with one data acquisition system and one set-up only, but the approach described is of a general nature that should be followed in a similar way for other situations. This means that the actual numbers presented here are not generally applicable, but the type of system verification and the type of validation models are. It is also clear that with multiple markers and different smoothing techniques, the actual accuracy might be different, but models as suggested in this paper should be used to show their effectiveness.

The three mechanical models have provided an assessment of a dynamic knee analysis system that relies on an optoelectric motion-tracking system. By representing specific aspects of the human knee, a tangible understanding of the joint mechanics was obtained, along with the measurement accuracy. Simple models provide a feel for the behaviour of the motion-tracking system as well as a verification of complex computer algorithms.

With camera–subject distances of 4–5 m, erroneous identification of LEDs warrants the separation of LEDs defining vectors, by more than 180 mm. The 180 mm obviously applies only for similar data collection systems in a similar set-up. It is important to note, however, that separation distances between any type of markers must be considered when a testing protocol is developed, as the chosen distances can greatly influence the accuracy. These distances can be reduced significantly by the use of multiple markers but, in many instances, it is desired to limit the number of wires in the trailing cable. The identification of LEDs is known to become poor as they are rotated away from the cameras (Scholz, 1989; Samuelson, 1987). However, by keeping the angle of presentation to within $\pm 15°$, this effect was negligible. The three relative rotations were calculated with acceptable accuracy, although they were not uniform between rotations. Varying the complexity of the motion from static to three degrees of freedom had no effect on the ability to calculate the rotation angles. Force calculations were accurate to within 8% and, by restricting the viewing range of the LEDs to $\pm 15\%$, the behaviour was consistent throughout the calibrated volume. The calibrated volume representing a cube with side lengths of 60 cm was sufficient as our interest was in the stance phase, but it could be necessary to calibrate a larger volume for other studies. Although the pendulum does not exploit the true three-dimensional nature of the mathematical model, it was able to indicate the absence of serious problems. More importantly, it provided a direct measure of the propagation of errors from data entry through algorithms involving differentiation and filtering. The homogeneity of the system within the calibrated volume implies that care need not be taken as to where in this volume the subject steps. This volume is large enough to ensure a natural gait.

While mechanical models can provide answers in the validation of a human evaluation tool, their use offers several limitations. Foremost is the fact that all modelling ignores skin motion. This is an assumption basic to the analysis of human movement. The motion of the underlying skeletal structures must be interpreted based on the surface markers (Chao, 1986). The interpretation becomes even more questionable with the introduction of probes used for three-dimensional analysis. Thus, it must be kept in mind that skin motion may exaggerate some of the errors found in this study. Human joints are much more complicated than mechanical joints; thus, the mechanical models do not replicate the dynamic behaviour of the true joint. For example, the knee is known to have a moving flexion axis, although we have assumed it to be fixed in the femur. These simplifications prevent a direct transferal of results from this study to the human motion study.

Motion-tracking systems of the type used in this study are dependent on their specific application. The accuracies delivered in this paper will apply to similar systems using comparable camera–subject configurations. However, the numerical results are not the essence of this thesis as it relates to other human motion studies. Rather, we recommend following similar procedures for the purpose of identifying what observable changes in the results are actually buried in system error. Each model supplied measures of practical relevance to a knee analysis study. They provided a direct output of the true measures evaluated, thus making an interpretation of the results, in a clinical context, possible. Knowing what observable changes are actually buried in system error allows the detection of statistically significant differences of selected measured parameters.

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REFERENCES

Andrews, J. G. (1984) On the specification of joint configuration and motions. *J. Biomechanics* **2,** 155–158.

Antonsson, E. K. and Mann, R. W. (1989) Automatic 6-DOF kinematic trajectory acquisition and analysis. *ASME J. Dynam. Syst. Neamt. Control* **III,** 31–39.

Chao, E. Y. S. (1986) Biomechanics of human gait. In *Frontiers in Biomechanics* (Edited by Schmid-Schoenbein, G. W., Woo S. L.-Y. and Zweitach, B. W.), pp. 225–244 Springer, New York.

Grood, E. S. and Suntay, W. J (1983) A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *J. Biomech. Engng* **105,** 136–144.

Li, J., Wyss, U. P. and Costigan, P. A. (1992) A new method for calculating bone-on-bone forces at the knee joint during gait. In *Proceedings of The Canadian Medical and Biological Engineering Society Conference*, Toronto, pp. 32–33.

Mann, R. W. and Antonsson, E. K. (1983) Gait analysis—precise, rapid, automatic, 3-D position and orientation kinematics and dynamics. *Bulletin of the Hospital for Joint Diseases Orthopaedic Institute* **XLIII,** 137–146.

Montgomery, D. C. (1984) *Design and Analysis of Experiments*. Wiley, New York.

Samuelson, B., *et al.* (1987) A device for the three-dimensional registration of human, movement. *J. Ergonomics* **30,** 1655–1670.

Scholz, J. P. (1989) Reliability and validity of the WATSMART three-dimensional optoelectric motion analysis system. *J. Phys. Therapy* **69,** 679–689.

Suntay, W. J. and Grood, E. S. (1978) A coordinate system for describing joint position. *J. Adv. Bioengng* 59–62.

Winter, D. A. (1979) *Biomechanics of Human Movement*. Wiley, New York.