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Repeatability of kinetic and kinematic measurements in gait studies of the lower limb amputee

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Abstract

During the last few years considerable attention has been given to the use of gait analysis as a tool for clinical use. The instrumentation for measurement of the kinetics and kinematics of human locomotion was originally designed for research use. Extension of its use into the clinical field calls for simplified methodology and clearly defined protocols with precise identification of the relevant parameters for the analysis. Force platforms, TV-computer and pylon transducer systems were used for collection of kinetic and kinematic data of five normal subjects, 10 below-knee, 10 above-knee and one hip disarticulation amputee. The repeatability tests showed significant differences in the measured parameters. These variations are attributed to the methodology of the analysis and the step to step variation of the subjects' gait. Differences in the degree of step to step variation between various amputee and normal subjects are quantified. In this presentation the capability of present day systems to perform repeatable gait measurements is discussed. A computational method for determination of representative measurements for the purposes of biomechanical evaluation and comparison as well as quantification of the degree of repeatability is described.

Introduction

The use of gait analysis for the assessment of several skeletal-neurological disorders, evaluation of the use of internal prostheses, measurement of effectiveness of orthotic devices, prescription of prosthetic components and fitting of lower limb prostheses is an inevitable and natural development of 50 years of research and development of instrumentation for the study of human locomotion undertaken at many centres throughout the world.

Since the beginning of contemporary gait studies by the University of California, commissioned in 1947, there has been an expansion in various parts of the world of development of instrumentation using modern technology and in planning long and short term programmes of research into the study of pathological and normal gait. Studies of biomechanics have allowed a new understanding of human locomotion particularly in respect of the forces developed at the joints of lower limbs and an indication of the proprioceptive feedback relating to position and velocity of the segments. The clinical application of gait analysis indicated the variability of the performance of normal and disabled subjects and highlighted the need for more repeatable and accurate measurements of kinetic and kinematic parameters. The use of such measurement facilities in the clinical situation assists the understanding of the gait process leading to identification and quantification of those variables which most accurately reflect the critical factors in gait of the disabled.

The most sophisticated methods of gait study have included force platforms for measurements of ground reaction forces, television/computer, infra red light sensing systems and cine photography systems in conjunction with passive and active body markers, for measurements of linear and angular displacements of limb segments. Additionally, studies have been performed to establish phasic muscular activity by utilizing EMG and metabolic energy cost by the use of respiratory gas analysis. Various computational techniques are used for data storage and reduction for the calculation of the position of joint centres and the loads developed there.

The accuracy and repeatability of the instrumentation used has been significantly improved. However the inaccuracies caused by the various assumptions made in calculations of

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joint centres and the repeatability of human movements and many other parameters have created a gap between the findings of the researchers in the field of human locomotion and the application of such findings in a clinical environment.

Repeatability of kinetic and kinematic measurements

Several factors may be identified which prevent the repeatable acquisition of gait data.

With regard to the instrumentation the leading commercial manufacturers of force platforms, Kistler and AMTI have reported the errors in signals from their load measuring devices as less than 2% with linearity errors less than 1% of full scale deflection. Similarly the TV/computer system marketed under the trade name "VICON" by Oxford Metrics for measurement of displacements is claimed to have an accuracy of 0.1% of the field of view. Other leading commercial systems in this field the CODA, SELSPOT and ELITE also claim similar or even better accuracy. Various individually designed instruments for load measurement such as the strain gauged pylon transducer for prosthetic load measurement and various kinds of kinematic measurement apparatus developed and used for gait analysis at research centres enjoy similar degrees of accuracy and repeatability.

The effects of computational analysis with fourth generation 16 and 32 bit computers have been quantified by several workers. Philippens (1981) examined the assumptions made in the analysis of data obtained from cameras in the study of motion and particularly the effect of accuracy of their placement and alignment. An error analysis was performed in the calibration, parallax correction and the use of three-dimensional direction cosine matrices for transformation of measured loads to various joint axes. Applying Philippens' analysis, a discrepancy of as little as two degrees in the alignment of camera and/or calibration frame, resulted in a typical calculation of antero-posterior hip joint bending moment being in error by 8% (The loads for these calculation were derived from a typical normal test performed as described in the methodology section).

Knowledge of the physical properties of the limb segments is a fundamental requirement in the study of human locomotion. The effects of various anatomical assumptions made in the calculation of the positions of joint centres have been reported by several workers. Various cadaveric studies have resulted in the development of constants for calculation of the positions of joint centres from the surface markers, and the calculation of mass and centre of gravity position of limb segments. A summary of the differences in the reported body segment parameters is given by Goh (1982). For the purposes of this presentation, the position of the hip joint centre was calculated with the aid of markers at the anterior superior iliac spines, sacrum, iliac crest, greater trochanter and the constants derived by Dempster (1955) and Ishai (1975). The estimated joint centre was then used for further calculation of antero-posterior bending moment at the hip. A difference of 12% was noted on using different derived constants.

There have been several publications quantifying the error produced by the use of passive and active body markers. These errors are in two groups. First the positioning of these markers and second the error caused by skin movement during the dynamic phase. In the latest reported work Andriacchi and Strickland (1985) have shown differences of 1.6 cm in positioning of an active body marker at the greater trochanter which is used for calculation of the hip joint centre. This resulted in a difference of 13.5% in the antero posterior bending moment at the hip. This effect varies at other levels (knee) and in other planes. Macleod of Oxford Metrics in a personal communication reported on the skin movements under passive markers during the swing and stance phases of normal subject locomotion. This effect has been illustrated by calculation of apparent differences of 3 to 4 cm in the length of the thigh segment.

Another major source of variation is the repeatability of the human gait itself. In quantification and identification of the individual source of variations it is important to measure the influence of each variable independently using controlled experimentation. Winter (1984) reported on the repeatability of the locomotion of normal subjects. A coefficient of variation as a measure of total variability was calculated. This parameter represented the root mean square of the standard deviation of the moment over the stride period divided by the mean of the absolute moment of force over the stride period. It was shown that on nine trial walks of one subject this coefficient of variability for ankle, knee and hip joint angle increased from nine to 10 and 19% respectively. Similarly for the axial load and shear force measurements, values of seven and 20% were reported. When moments at ankle, knee and hip in the antero-posterior plane were considered, the coefficient of variability increased from 22 (ankle) to 67 (knee) and 72% (hip) respectively.

The use of gait analysis has been reported in the evaluation of the use of orthoses in cerebral palsy children (Gage, 1983; Meadows, 1984), on evaluation of hip joint replacement (Kelly et al, 1983), on the influence of alignment on amputee gait (Zahedi, et al, 1987) and many other applications which require evaluation and comparison of individual gait patterns. However very few attempts have been made to describe the repeatability of measurements from one test subject. For the purposes of such comparison it is necessary to quantify the repeatability due to the method of measurements and step to step variation before attempting any biomechanical comparisons.

To show the step to step variations in normal subjects requires facilities such as long force platforms for measurements of ground reaction forces or devices such as load measuring devices inserted in the shoes or attached on the outside. Mobile camera systems or other systems or goniometers are required to acquire data accurately over several steps. Moreover the independent influence of anatomical assumptions, computational analysis, body marker positioning and skin movements have to be controlled and quantified. At present most clinical gait analysis laboratories have only conventional force platforms and kinematic measuring systems capable of handling data acquisition for one single step. It is therefore necessary to quantify the run to run variation from the selected measured steps. With regard to the prosthesis of a lower limb amputee, the effect of the anatomical assumptions, the positioning of the markers and skin movement do not apply. Further, the strain gauged pylon transducer incorporated in the shank of the prosthesis allows the monitoring of the three orthogonal forces and moments for an infinite number of successive steps constrained only by the testing environment. Furthermore, by measuring the alignment of the prosthesis it is possible to establish the position of the ankle, knee and hip relative to the pylon transducer and thus calculate the loadings at various levels. Thus it is possible to quantify the actual degree of step to step variation independently.

Methodology

Figure 1 shows the set up of the biomechanics laboratory at the Bioengineering Unit,

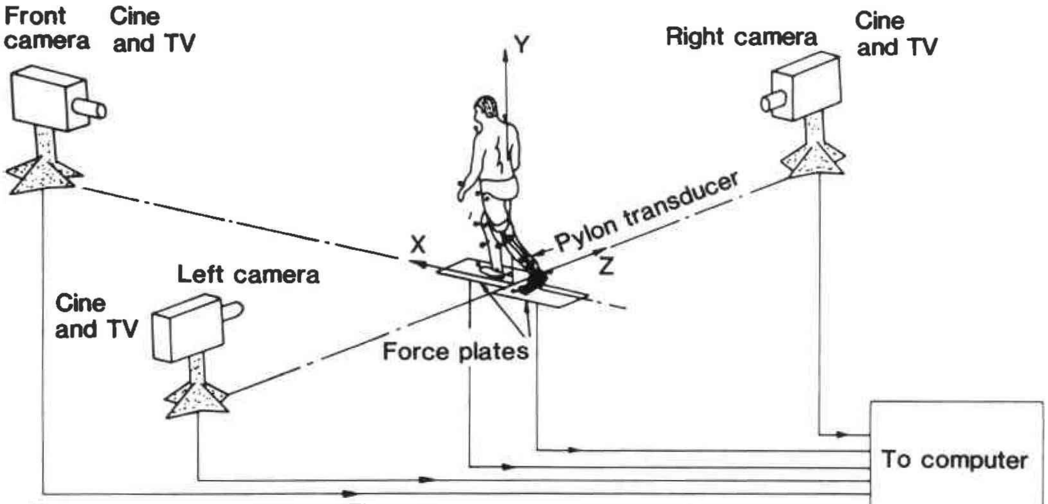

Fig. 1. The biomechanics laboratory.

University of Strathclyde, Glasgow, Scotland. The Kistler force platforms, and the Strathclyde Television Computer system were used for measurements of kinetic and kinematic parameters. Additionally the strain gauged short pylon transducer (Berme et al, 1976) as seen in Figure 2 incorporated in the shank of the prosthesis was used. A single axis goniometer measured the flexion angle of the prosthetic uniaxial knee joint. This angle in conjunction with alignment measurements, defining the relative geometrical position of the socket, the knee (as appropriate) and the foot of the prosthesis allowed the positions of the hip, knee and ankle relative to the transducer to be calculated. The Unit's PDP 11/34 computer was used for all data acquisition and data reduction.

Four categories of test subjects were considered as shown in Table 1.

With the exception of the last category of these subjects, the remainder were selected for participation in the programme of the study of alignment of lower limb prostheses undertaken at the Bioengineering Unit (Zahedi et al, 1986; Zahedi et al, 1987). The hip disarticulation subject was used for the study of the biomechanics of hip disarticulation. (Solomonidis et al, 1977).

As far as possible, standard prostheses and means of measurement were used to allow comparison of the results with those of other investigators. Otto Bock modular prostheses were used to allow easy change of alignment. The mass of the prostheses excluding the cosmetic covering but with the addition of the 500 g transducer was on average within ± 10% of the mass of the definitive prosthesis to which the amputees were accustomed. The prostheses were fitted to the amputees and dynamically aligned by the prosthetist. In each case two other prosthetists verified the fit of the socket and the alignment. Questionnaires completed by them were later correlated. An unsatisfactory collective rating of a fitting

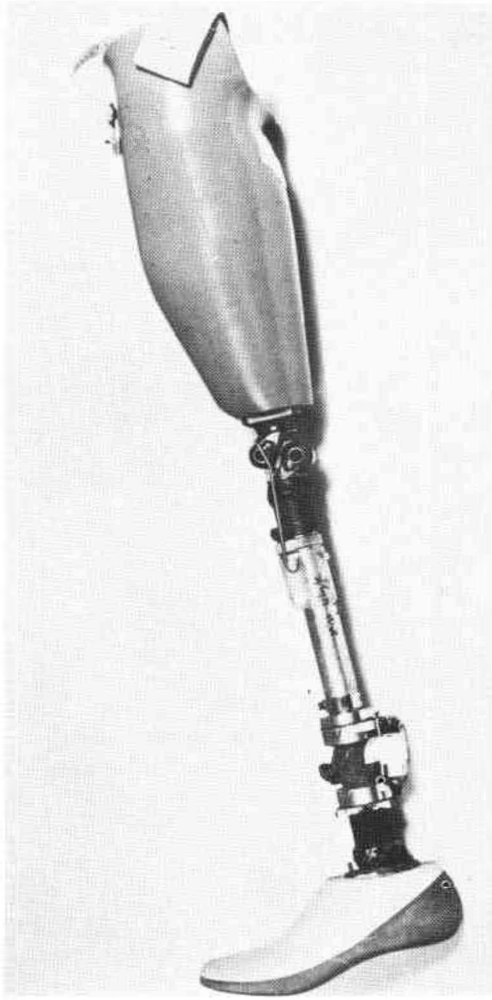

Fig. 2. Pylon transducer incorporated in the shank of an AK prosthesis.

Table 1. Details of subjects.

Subjects		Age		Years since amputation		Activity level*	
Type	No.	Mean	S.D.	Mean	S.D.	Mean	S.D.
Normal	5	36.2	± 9.51	—	—	—	—
Above-knee	10	46.1	±11.80	13.3	8.2	31.10	±13.43
Below-knee	10	55.33	±10.66	17.4	±10.5	37.5	±10.28
Hip Disarticulation	1	21	—	2	—	17	—

*Activity level determined according to Day (1981) (above +30 very active, below −40 inactive).

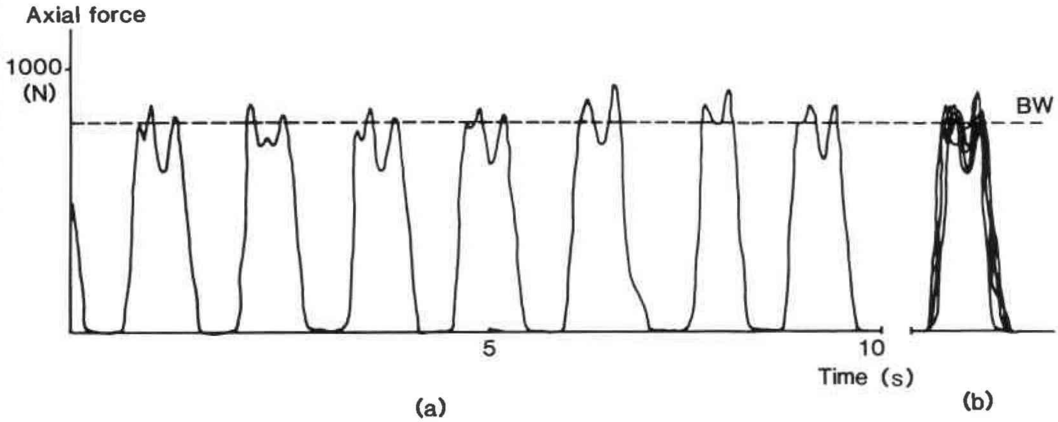

Fig. 3a. Prosthesis axial force for 9 successive steps by BK amputee; 3b, superimposition of cycles shown in 3a.

resulted in elimination of that test. A time of approximately one hour was given to the amputee to get used to the prosthesis and surroundings.

The force plate/TV data acquisition system was used for measurement of single steps of several successive runs for the normals and the sound and prosthetic side of the amputee subjects. The pylon transducer was used for prosthetic load measurement of successive steps on each run of the amputee subjects inside the laboratory. A series of Fortran programs allowed the sampling, filtering and calibration of data and transformation of loads to the ankle, knee and hip levels and the display of the results.

As the results from several successive steps and the selected steps from various runs showed different periods for stance and swing phase timing, it was not possible to simply average the data by the simple process of adding the signals without significantly distorting the picture. Simple normalization to the minimum time base also resulted in distortion of the amplitude of the signal (since the variation in the cycle time could be as much as 25%). Thus a technique was developed which firstly, automatically selected individual steps with the start, end and period of each step determined. Then a Fourier analysis technique was used to normalize the waveforms of different periods. Once the harmonics were determined, a new data sequence was generated based on a normalized period, without distorting the

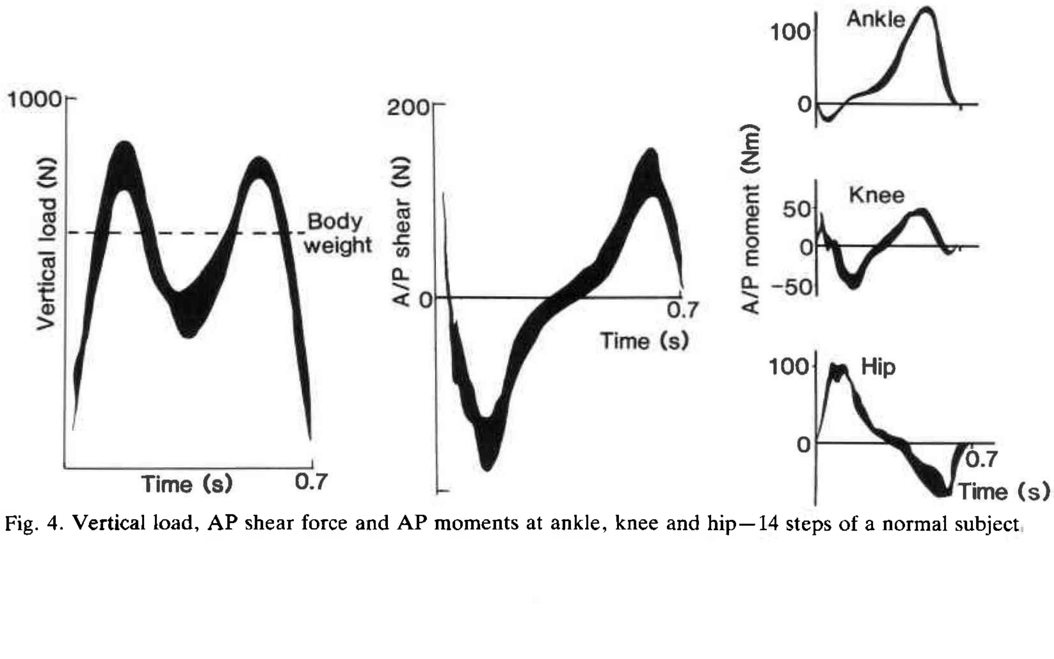

Fig. 4. Vertical load, AP shear force and AP moments at ankle, knee and hip—14 steps of a normal subject.

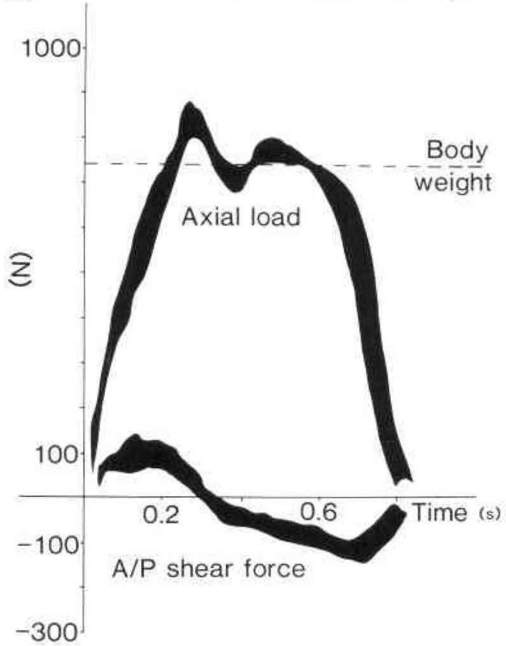

Fig. 5. Axial load and shear force of 14 steps by an AK amputee.

amplitude of the signal. The data can now be averaged for all successive steps of a run or the selected individual steps of several runs. The standard deviation is also quantified. The plot of the averaged data as the representative step for analysis and the standard deviation plot as the representation of the degree of repeatability is produced. The superimposition of the individual signals for each step develops an envelope which produces another form of representation of the degree of the repeatability. Figure 3 shows the axial loading of the prosthesis for a below-knee amputee: nine successive steps are presented, measured using the pylon transducer. The degree of step to step variation is shown by the thickness of the superimposed envelope of the individual signals.

Results

Figure 4 shows the curves of variation with time of load actions on the leg in the sagittal plane for a typical normal subject walking 14 times over the force platform. The composite curves are formed by superimposing the values for the fifth step in each of the 14 tests. The thickness of the envelope represents the variation between tests. These variations are due to step to step variation and skin movements under markers. Figure 5 represents the axial and shear force in the pylon frame of reference for an amputee using the same

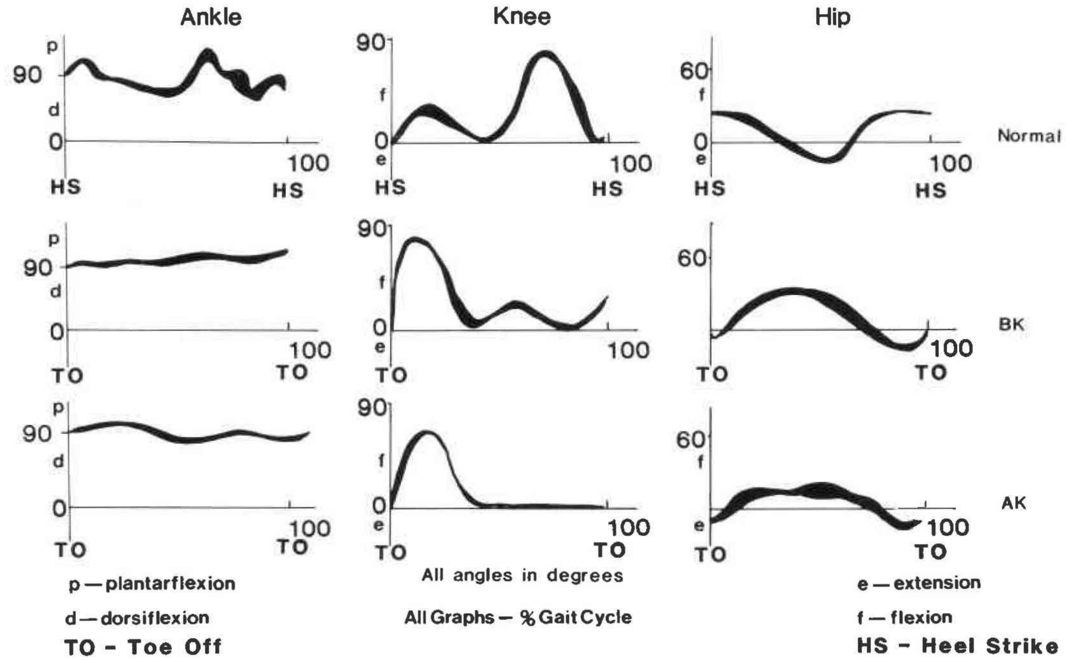

Fig. 6. Joint angle variation with time for 14 steps of a normal subject, an AK amputee and a BK amputee.

prosthesis over 14 steps. There are no skin movements and no errors due to positioning of the markers as the joints are mechanical and the socket reference land marks, used for measurement of alignment (Zahedi et al, 1986), are permanently marked. However there is variation corresponding to the step to step variability. The repeatability of kinematic data is illustrated in Figures 6 and 7, which show joint angle plotted against time and angle versus angle diagrams, for a normal subject, a below-knee and an above-knee amputee. Figure 8 shows the forces and moments measured by the pylon transducer incorporated in the shank of an above-knee prostheses for 60 successive steps on six runs. These signals are normalized and averaged using the previously described technique (Fig. 9). The standard deviation represents purely the degree of step to step variation (Fig. 10). The repeatability of

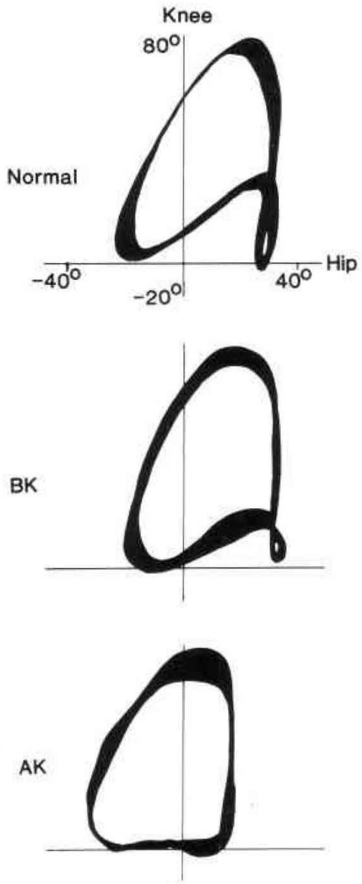

Fig. 7. Hip v knee angle diagram for 14 steps by a normal subject, an AK amputee and a BK amputee.

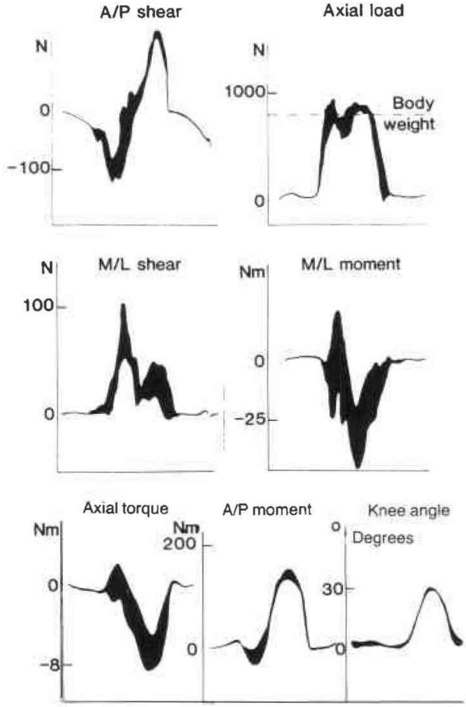

Fig. 8. Forces and moments of 60 steps by an AK amputee at pylon level.

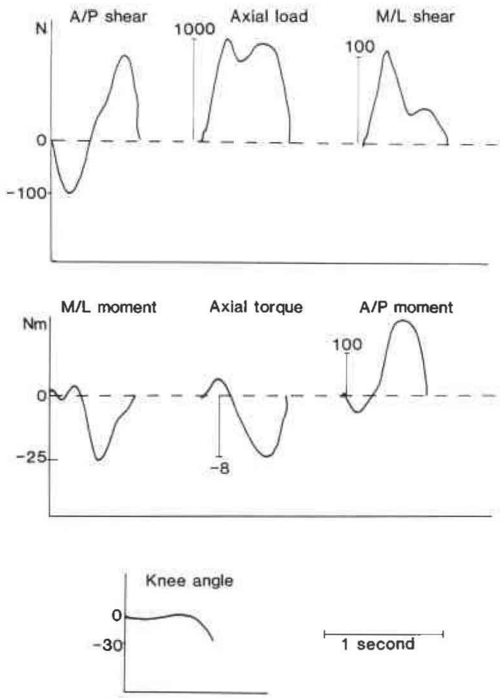

Fig. 9. Average of data at ankle level in shank frame of reference.

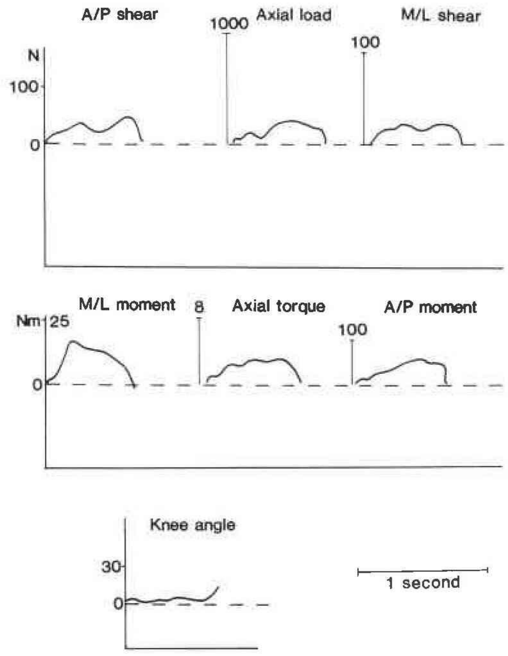

Fig. 10. Standard deviations of data (positive values only are shown).

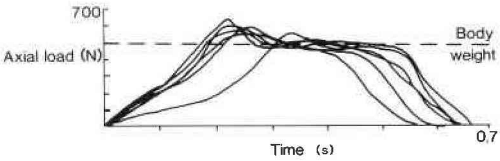

Fig. 11. Axial load in the prosthesis of a hip disarticulation amputee measured during 6 steps.

the axial load measurement for a normal subject is compared with the below-knee, above-knee and hip disarticulation data from Solomonidis et al (1977) in Figure 11.

The prosthesis for an above-knee amputee was dynamically aligned by three different prosthetists. Both the amputee and prosthetists were satisfied with the alignment and amputee's gait. Figure 12 shows the axial loading, the antero posterior, and mediolateral external bending moments at the ankle level. Figure 13 shows the comparison of the representative signal taking into account the step to step variation and the standard deviation graphs as the value for the degree of step to step variation for each condition.

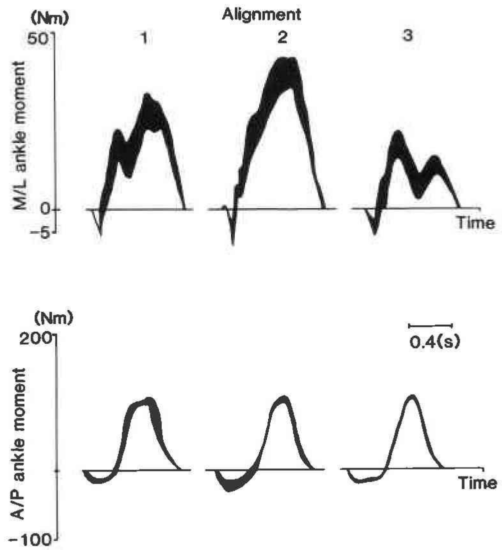

Fig. 12. Repeatability of AP and ML bending moment on 1 AK amputee with 3 different acceptable alignment configurations.

Discussion

The repeatability of the kinetic and kinematic measurements in the study of gait is determined by the errors produced in the instrumentation, techniques of data acquisition and the repeatability of the human gait. The errors caused by anatomical assumptions for location of the joint centres can be reduced by subjective selection of the reported constants (body segment parameters) and direct measurement where possible. The body segment parameters derived by using subjective averaging (Goh, 1982) provided better results. Where X-rays are possible direct anatomical measurements can be made reducing the errors

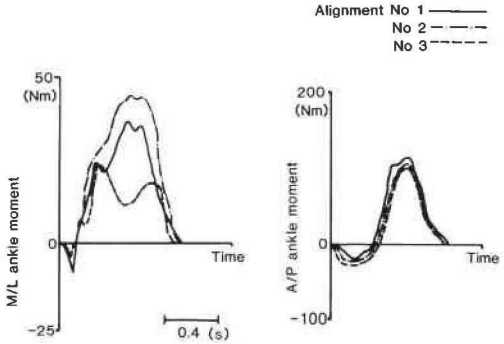

Fig. 13. Representative moments for 3 acceptable alignments.

even further (provided parallax is corrected). The errors caused by computational analysis to some extent have been overcome by the use of space calibration techniques as described by Marzan and Karara (1975) which take account of camera alignment and lens distortion by spatial calibration. The error in body marker positioning can be reduced to a very insignificant level by using small passive markers of 10 mm diameter assembled on a flexible base which can be accurately and repeatedly mounted on the anatomical land mark. The mass of a passive marker can be made very small (approximately 1g) made from solid foam. Mounting sites with least soft tissue under the skin can be selected. This method may require more computation of static and dynamic analysis than simple markers over joint centres. However by combining the above criteria, faster sampling of data (increase to 100 Hz from 50 Hz) and introduction of an averaging technique during the sorting of each marker position, the errors are reduced significantly.

The main source for the inability to produce repeatable results is therefore demonstrated to be the step to step variation within several successive steps and the variation between the selected step of several runs. It was found that the best representative step for analysis, in a sufficiently long gait laboratory (15 m) which covers 15 steps of a normal subject is step number seven for force plate/TV data acquisition. Using video recording of the subjects' locomotion, the first and last three steps were found to be visually different from the rest of the steps.

The quantification of the repeatability of individual subjects also allowed the overall assessment of variation from one subject to another. Therefore it becomes important to have a measure of the repeatability for every subject tested, especially when biomechanical comparisons are made, or long term biomechanical changes in subjects' gait are to be monitored. This is crucial if small changes such as small differences between equally acceptable alignments in a lower limb amputee are being considered. It has been a conventional belief that there is a single optimum alignment for a prosthesis corresponding to 'best' gait performance. It has however been shown that, under the present method of prosthetic fitting the amputee could be satisfied with more than one alignment configuration (Zahedi et al, 1986). As can be seen, small differences in alignment influence the degree of repeatability as well as the pattern of the actual load actions, although the subject and the prosthetist are not aware of such differences and are satisfied with all alignments. From biomechanical considerations, alignment No. 3, which has also the least variation in loads from step to step in antero posterior bending moment and axial loading parameters seems better and nearer to the optimum condition. It is suspected that the amount of step to step variation is dependent on the degree of control during the gait. As equilibrium has to be maintained in each step, the reaction forces and moments measured indicate the required balance of forces and moments for spatial stability and equilibrium. Thus the subject has to correct for changes in forces, position and velocity using the proprioceptive feedback control for each step. This is seen as a variation from one step to another in the measured parameters. This could also be seen in variation of step to step changes in the normals and in further comparison with amputee subjects. A loss of a limb will reduce the degree of control of the musculo-skeletal structure and reduce proprioceptive feedback for the ambulation of the subject. Further this loss of control is greater in an above-knee amputee and even greater still in a hip disarticulation patient than in the below-knee amputee. The variation from step to step can be visually detected in the gait of a hip disarticulation amputee and to a lesser extent in above-knee amputees. However, visually it is very difficult to detect the step to step variation in all active below-knee amputees and it is almost impossible to detect this variation in normal subjects. Thus it can be concluded that for optimization of gait in an amputee it is necessary to quantify the degree of repeatability and to use the true representative signal for the biomechanical analysis.

Conclusion

In the use of gait analysis for the study of normal and pathological locomotion and for the purposes of clinical comparison, it is first necessary to quantify the degree of the repeatability due to the method of measurement and step to step variation, before attempting biomechanical comparison.

The degree of step to step variation varies for different normal subjects. This degree of step to step variation increases for the amputee population compared with normals. It appears that within the amputee population it increases further as the level of amputation becomes more proximal.

The degree of repeatability of kinetic and kinematic parameters measured at various levels increases with proximal increase in level from the ground.

A method is now available for quantification of step to step variation and reduction of the data to a representative signal for the purposes of biomechanical comparison.

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