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**VALIDATION OF AN INVERSE DYNAMICS METHOD
TO PREDICT JOINT KINETICS IN THE ABSENCE
OF DYNAMOMETRY**

by

John M. Barden

Thesis submitted to
the School of Graduate Studies and Research
in partial fulfilment of the requirements for
the Master of Science degree in Kinanthropology

University of Ottawa



John M. Barden, Ottawa, Canada, 1993



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DEDICATION

This thesis is dedicated to my parents, John and Sylvia, who have taught me many things, among them that learning can be and should be fun. It is also dedicated to my brother Edwyn who helped me by participating as a subject and my wife Rhonda, who gave her time, understanding and patience in many ways and on many separate occasions throughout the completion of this project. I love you and thank you all.

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ABSTRACT

When investigating the joint kinetics of human locomotion, ground reaction forces are typically measured directly using a force platform. In many different skills and movement environments, the utilization of a force platform to collect external force data is not possible. These particular movement situations require an indirect method to estimate joint kinetic variables if they are to be analyzed. This study investigated whether or not an eleven segment, complete body model could be used to accurately predict the net joint reaction forces and moments of force from cinefilm and body segment parameter data, using force platform calculated net joint reaction forces and moments of force as the criterion for validation. The body was modelled as a linked system of pin-connected rigid bodies and was restricted to the sagittal plane analysis of single support locomotions. The movements analyzed for three subjects were jogging, running, the acceleration phase of sprinting, the long jump takeoff and a running front somersault into a forward roll. Displacement and ground reaction force data were both sampled at 100 Hz and low-pass filtered at cutoff frequencies of 6 and 10 Hz, respectively. The film predicted, support limb net joint reaction forces and moments of force were compared to the force platform calculated results using RMSE and Pearson product moment correlation statistics to quantify phase and magnitude variations. It was found that vertical net joint reaction forces can be estimated from displacement data using the inverse dynamics approach and a complete body segment model. Limited success was attained in estimating the horizontal net joint reaction forces. No consistent results were obtained in estimating the net joint moments of force. Further refinements are necessary if net joint moments are to be predicted accurately using this particular method.

INTRODUCTION

The human movement aspect of biomechanical research is beneficial to a wide variety of health science professions because of its ability to quantify the variables associated with human motion. Physicians, physiotherapists, chiropractors, occupational therapists and athletic coaches represent a few of the professionals who are able to benefit from biomechanical research.

An important aspect of human movement biomechanics is to provide quantitative information about the performance of major muscle groups in a given activity. A primary source of this information involves calculating the various net joint reaction forces and moments of force. These variables provide a powerful analytical tool, with which it is possible to gain insight into the net summation of all muscle activity at each joint. Net joint reaction force and moment time histories are calculated using linked segment rigid body mechanics. This approach requires a full kinematic description of the motion, accurate anthropometric parameters and the magnitude and direction of the ground reaction forces. Typically, ground reaction forces are directly measured using a force platform or other type of force transducer.

In many human movement situations, the utilization of a force platform to collect the ground reaction force data is not possible. A number of difficulties are associated with the use of force platforms which has limited their application in rehabilitation centres, factories, athletic clubs and many sporting environments. Some of the limitations associated with the use of force platforms are: 1) they are expensive to purchase, 2) they must be properly

secured (i.e., built into the floor) to ensure adequate damping, 3) they can sometimes be heavy and difficult to transport, 4) they usually have a surface area that is quite small, sometimes making contact difficult for subjects with abnormal gait patterns (e.g., handicapped and paraplegic patients) and 5) they can be similarly restrictive in sporting events if athletes have to adjust their pace or alter their motion to be able to make contact with the surface of the plate. As a result, force platforms are not often used in testing environments outside of the laboratory or clinic.

This has limited the ability of biomechanics to perform kinetic analyses of human movement in the majority of field study situations. Many sporting activities in particular, suffer because they are not conducive to being performed in a laboratory environment, and could benefit from field study testing. Examples of such sporting activities are figure skating, cross country skiing and springboard diving, in addition to many competition situations where it is impossible to conduct a kinetic analysis of the motion using a force platform. It is unfortunate that these and other activities like them should be neglected when knowledge of the joint kinetics could improve the coaching and performance of these skills.

The calculation of net joint moments requires knowledge of the joint reaction forces, which in turn are influenced primarily by the ground reaction force components. Although it is desirable to measure the ground reaction force components directly, they can be predicted from kinematic and inertial property data using the procedure known as inverse dynamics (Winter, 1990). Some work has been done on predicting (or deriving) the horizontal and/or vertical (primarily) ground reaction force components from double differentiated displacement data (Smith, 1975; Thornton-Trump and Daher, 1975; Robertson, 1980;

Patriarco et al., 1981; Zarrugh, 1981; Miller and Nissinen, 1987; Sanders et al., 1991; Bobbert et al., 1991). Of the researchers who studied displacement predicted ground reaction forces, only Sanders et al. (1991) quantified the difference between the predicted and directly measured (force platform) vertical force curves. Only a few of those who predicted ground reaction forces also predicted joint moments and compared these with joint moment histories calculated using force platform information (Thornton-Trump and Daher, 1975; Robertson, 1980; Patriarco et al., 1981; Zarrugh, 1981).

A variety of linked segment models have been used in prior investigations, but it remains to be seen if accurate joint kinetics can be derived from displacement data using a complete body segment model for movements other than walking. For this reason, it was decided that it would be worthwhile to investigate whether or not a complete, linked segment body model could be used to predict accurate joint kinetics, using force platform calculated joint kinetics as the criterion for validation. Robertson (1980) was successful in predicting accurate sagittal plane net joint moments for the ankle and knee in normal level walking, but reported unsatisfactory results for the hip. Of the studies which have compared displacement predicted and force platform calculated joint moments, those which lacked their own force platform validations compared their findings to the results published in other studies (Thornton-Trump and Daher, 1975; Zarrugh, 1981). Unfortunately, differences in body segment parameter data, link segment body models and data collection and processing procedures make accurate comparison of these results somewhat dubious.

To summarize, the problems and additional costs and restrictions associated with dynamometry, along with the need for field study research in many situations, make it

desirable to attempt to utilize indirect methods to estimate complete joint kinetics from the already necessary displacement and body segment parameter data. An attempt to validate the ability of a total body, linked segment model to predict sagittal plane joint kinetics is the first step in the development and refinement of a successful method.

The purpose of this study was to predict the net joint reaction forces and moments of force from cinefilm and body segment parameter data, and to validate these estimates by comparing them with the results calculated using force platform determined ground reaction forces. The present study modelled the human body as a system of one-degree of freedom, pin-connected rigid bodies and was restricted to the sagittal plane analysis of single support human locomotions.

METHODS

Data collection procedures

Three subjects (two male and one female) performed a series of single support, primarily sagittal plane motions. The motions analyzed for all three subjects were jogging, running and the acceleration phase of sprint running. The takeoff phase of the long jump and a running front somersault into a forward roll were also analyzed for both male subjects. These particular motions were chosen for analysis because they represented a variety of single support human movements, included a number of different speeds and were appropriate for potential sport biomechanics applications. Two trials of each motion were

analyzed for each subject, following numerous practice trials to allow them to become proficient in making contact with the inlaid force platform without having to change pace or alter their motion. In the front somersault trials the two lower limbs were assumed to be bilaterally symmetric and were modelled as a single limb. Subject preparation included the placement of markers on selected anatomical locations along with the measurement of segment lengths and total body mass. Markers were placed on the skin at the following locations according to segment length definitions specified by Dempster (1955) and Winter (1990): the shoulder, elbow, wrist, hip, knee, ankle, heel, ball of foot and toes for both limbs, in addition to the ear and iliac crest.

Displacement data were collected at 100 Hz using a LOCAM 16mm cinecamera with a 12-120 mm Angenieux zoom lens. Cinefilm was chosen as the displacement data collection method because it was anticipated that it or video would most likely be the data collection choice for field study applications. In addition, film analysis is easily used outside of the laboratory, provides a permanent record of the trial and does not encumber or modify the movement of the subjects being tested. The camera was situated perpendicular to the x -axis of the force plate and laboratory runway at a distance of 7 m so that the subject and force plate area filled the entire field of view. The filming protocol consisted of: 1) the filming of a grid board (with control points 25x25 cm apart) in the plane of motion to scale the data, 2) the filming of static profiles (left and right sides) of the subjects and 3) the filming of the required trials. The grid board was used to obtain a set of control points to estimate the external camera parameters (i.e., position and attitude) and the coefficients of a fractional linear transformation to translate raw cinefilm coordinates into a true sagittal reference plane.

Ground reaction forces and centre of pressure distances were recorded with a six quantity, piezoelectric force plate manufactured by Kistler Instrumente AG of Winterthur, Switzerland (Kistler force plate type 9261A). Force platform signals were sampled at 100 Hz and processed on a Data General S/20 microEclipse minicomputer with A/D conversion capability. An LED counter connected to the data collection computer was placed in the camera field of view during filming. The counter was triggered the instant the force platform began to collect data. This enabled the cinefilm and force platform recordings to be properly synchronized.

Data analysis

The developed cinefilm was digitized with a Hewlett-Packard (HP 9874) digitizer and then processed, along with the raw force data, using a new, specially modified version (University of Ottawa) of the BIOMECH (University of Waterloo) analysis package. A minimum of five extra film frames (in most cases ten) were digitized prior to heel-strike and following toe-off to eliminate filter boundary effects. The static profile views of the subjects were used to construct transparent overlays of the subjects' segments. These transparencies allowed markers on the far side of the body, and on the hip marker closest to the camera, to be estimated and digitized when they became hidden or partly obscured by the other body segments. The transparencies were also used to minimize trunk segment kinematic error by using the hip, shoulder and iliac crest marker overlay positions to maintain a stable trunk segment length when digitizing. Robertson (1980) and Miller and Nissinen (1987) have

identified the trunk segment as being a significant source of potential error in the determination of whole body kinematics.

The displacement and raw force data were filtered with a 2nd order, zero lag, low-pass Butterworth digital filter using cutoff frequencies of 6 and 10 Hz, respectively. These cutoffs were chosen to maintain high signal-to-noise ratios based on sampling theorem guidelines and the proposed 6 Hz maximum frequency of human movement (Winter, 1990). The effect of varying the displacement data cutoff frequency on the predicted joint kinetics, was examined for cutoff frequencies between 15 and 2 Hz.

A subroutine of the BIOMECH package (KINEMATIC program) employs finite difference equations to compute both the linear and angular velocities and accelerations of the segment centres of gravity. Pezzack et al. (1977) have shown that the process of digitally filtering cinefilm coordinate data and then applying finite differentiation, is an appropriate solution for obtaining segmental kinematics. Body segment parameters were also computed with the BIOMECH package, using the anthropometric proportions of Dempster (1955) and Plagenhoef (1971) for the male and female subjects, respectively.

Segment body model

Two linked segment rigid body models were used in this study; one for calculating the joint kinetics of the support limb using the force plate and film determined ground reaction and inertial forces, and the other to predict the joint kinetics of all four limbs from the cinefilm obtained displacement data. The force platform model was a four segment

model consisting of the trunk, thigh, lower leg and foot segments of the support limb. The film prediction model was an eleven segment model consisting of the right and left feet, lower legs, thighs, arms, forearms and a head, neck and trunk segment. A stick figure representation of this model for one of the trials is depicted in Figure 1. In the interest of examining the effect of the different limbs on the model's ability to predict accurate joint kinetics, one trial was analyzed using the model described above excluding both of the arms (i.e., trunk, swing and support legs only) and excluding the swing leg (i.e., trunk, arms and

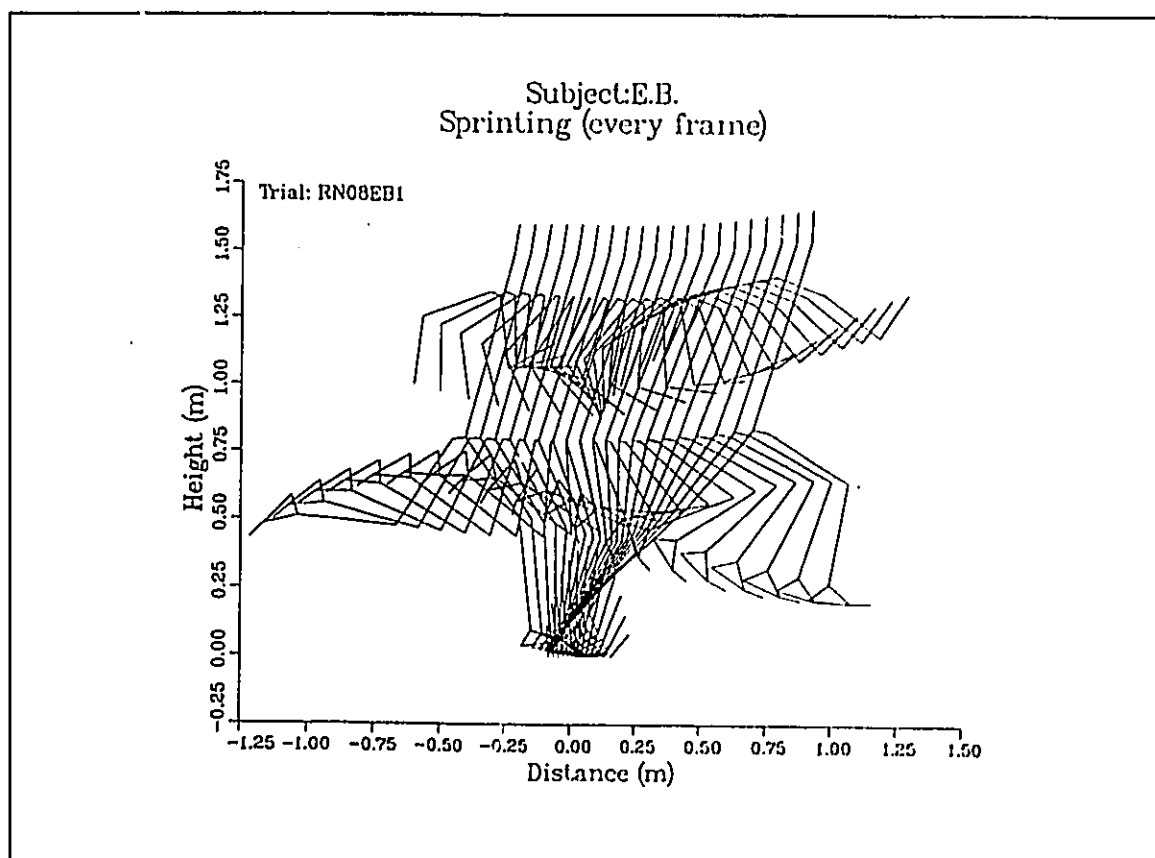


Fig. 1. Stick figure representation of film prediction linked segment model.

support leg only).

The net joint reaction forces and moments were calculated with the University of Ottawa's own adapted version of the BIOMECH package, specially written for this thesis, using standard linked segment rigid body mechanics. In the case of the joint kinetics for the force platform model, the net joint forces and moments were calculated by successive applications of equations of dynamic equilibrium to the distal-most and then more proximal segments knowing the: 1) segment kinematics, 2) inertial properties and 3) all externally applied forces.

A slightly different approach was taken in predicting (or estimating) the net joint reaction forces and moments from the film displacement data. In contrast to beginning the procedure at the proximal end of the foot, the method began at the proximal end of the right and left forearms. The same progression of calculating first the distal and then the proximal net forces and moments of a segment still applied. The joint kinetics in this case are calculated knowing the: 1) segment kinematics and 2) inertial properties. To briefly illustrate, the proximal net forces and moments of the forearm (i.e., at the elbow joint) were determined, reversed and applied as the distal net forces and moments of the upper arm. This procedure was employed for both arms and the swing leg, permitting the calculation of the net forces and moments at the support hip joint on the trunk. The net forces and moments at the knee and ankle joints of the support leg were subsequently determined in a proximal to distal progression.

Validation of the predicted joint kinetics

To be able to validate the film displacement method used to predict the net joint reaction forces and moments of force, both the predicted and criterion (force platform measured) quantities were plotted on the same axes. The root mean square error (RMSE) between the criterion and predicted force and moment histories were calculated to give an indication of the error in magnitude associated with the predicted quantity. A RMSE of 10% of the maximum range or lower was considered to represent a sufficiently accurate portrayal of the magnitude of the criterion force or moment time history, as predicted by the film displacement method. Pearson correlation coefficients were calculated to quantify the degree to which the predicted force or moment histories were "in phase" with the criterion quantities. This comparison is important because it provides a measure of the film prediction method's ability to accurately reproduce the pattern of the motion, as indicated by the changing net joint force and moment time histories.

RESULTS

Comparison of measured and predicted force curves

Figure 2 shows a comparison of the force platform calculated and film predicted net joint reaction forces in jogging, for the ankle, knee and hip joints. The velocity for the total body centre of mass at ipsilateral foot-strike was 3.8 m/s. The predicted force curves show

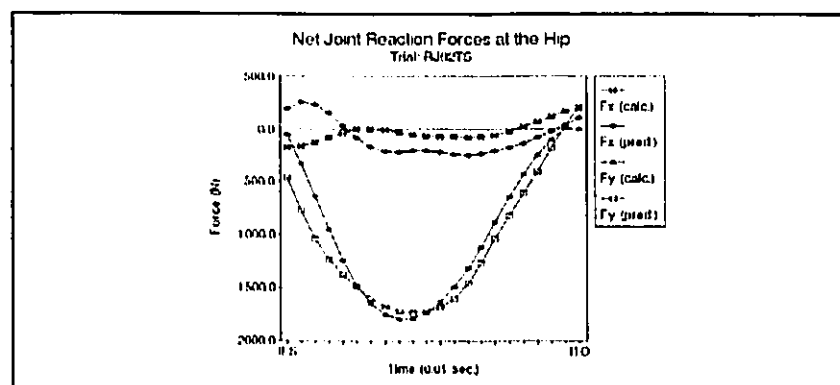
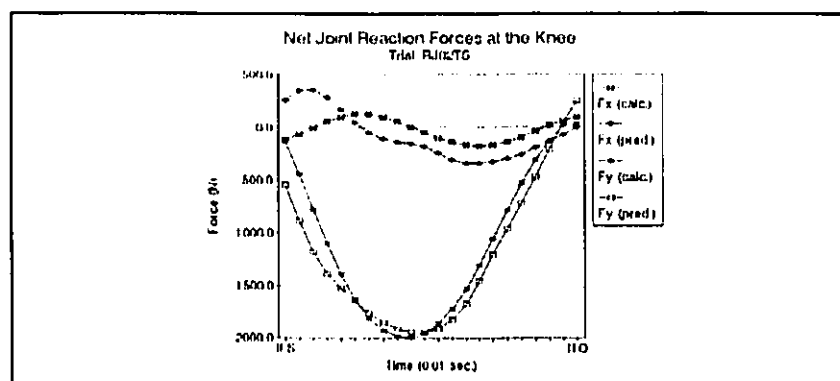
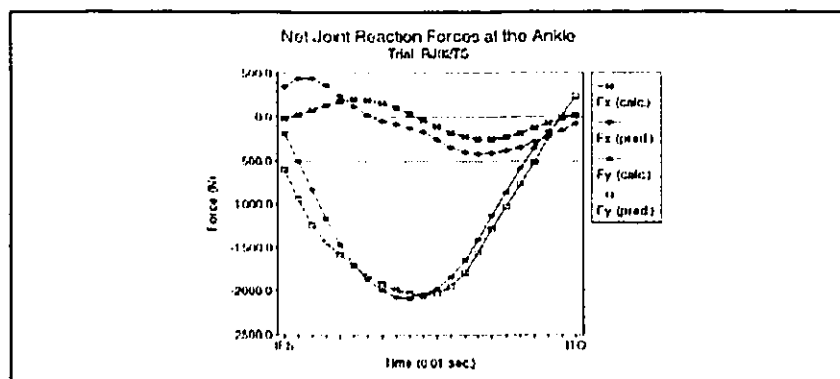


Fig. 2. Examples of force platform calculated and film predicted net joint reaction forces for the ankle, knee and hip when jogging ($v = 3.8$ m/s).

good agreement with the directly measured curves for both F_x and F_y . The predicted F_x curve has a small peak occurring at foot-strike which is inconsistent with the criterion, and also slightly overestimates in the negative direction. This overestimation of F_x was common throughout all of the trials to varying degrees. For this particular trial, the predicted curves mimic the measured criteria for F_x and F_y quite well.

Figures 3 and 4 compare the predicted and calculated results at the knee for running and the long jump takeoff. F_y is predicted well for both trials, somewhat more so for running. F_x is also predicted well, but again overestimates the calculated curve particularly in the first and last thirds of the support phase. Due to space limitations it was not possible to show all of the subjects' trials, or all of the results at each joint for the trials that were selected. In total, more than 150 film predicted and force platform calculated force and moment graphs were analyzed. The trials that were selected were chosen because they most accurately represented the results of this research. In addition, it was not necessary to present net joint reaction force graphs at each joint for all of the selected trials, because the results were similar (visually, not statistically) for the three joints within each trial, as evidenced by the results in Fig. 2.

Comparison of measured and predicted moment curves

Figure 5 shows an example of the predicted and calculated net moments of force for the ankle, knee and hip joints when jogging. It was obvious from these results that the film prediction method did not work well in estimating the net joint moments of force.

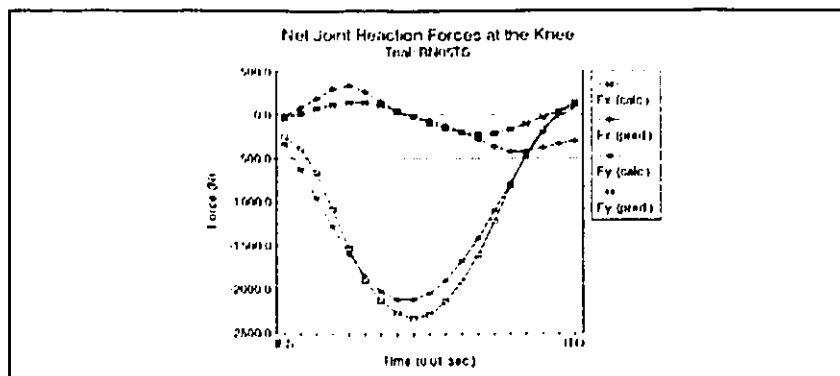


Fig. 3. Comparison of film predicted and force platform calculated reaction forces at the knee when running ($v = 4.3$ m/s).

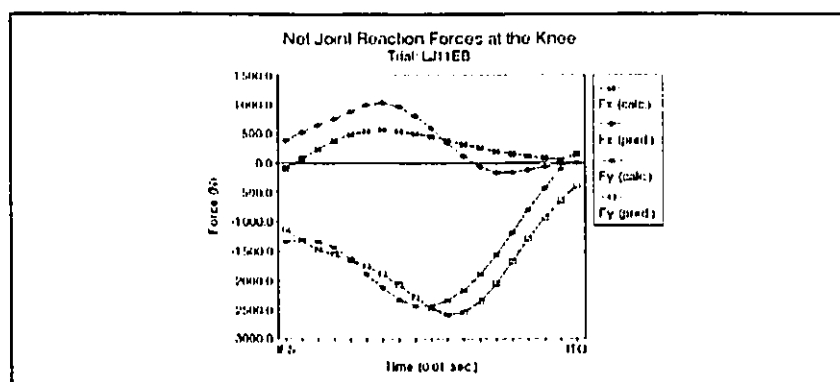


Fig. 4. Comparison of predicted and calculated reaction forces at the knee in the long jump takeoff ($v = 4.9$ m/s).

Interestingly, the predicted curve was much better for the knee joint than it was for the hip or ankle joints. In a number of instances, the predicted moment curves at the knee were quite similar in shape to the calculated (i.e., criterion) curve (see Fig. 6). While this initially appeared encouraging, the overall results were very unreliable. Although not immediately apparent, if the predicted curves at the ankle and knee shown in Fig. 5 are compared on a

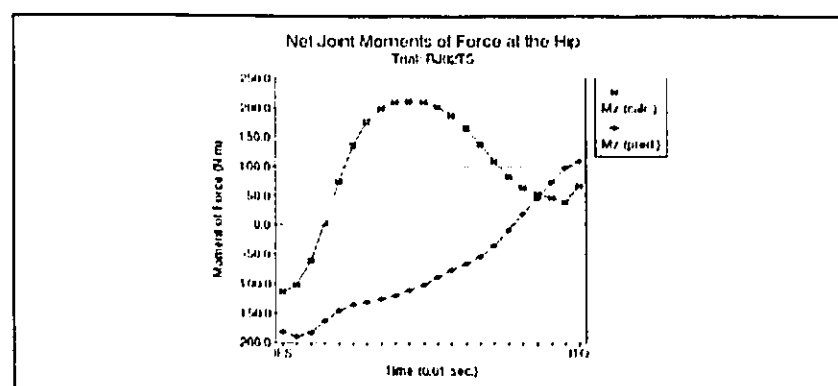
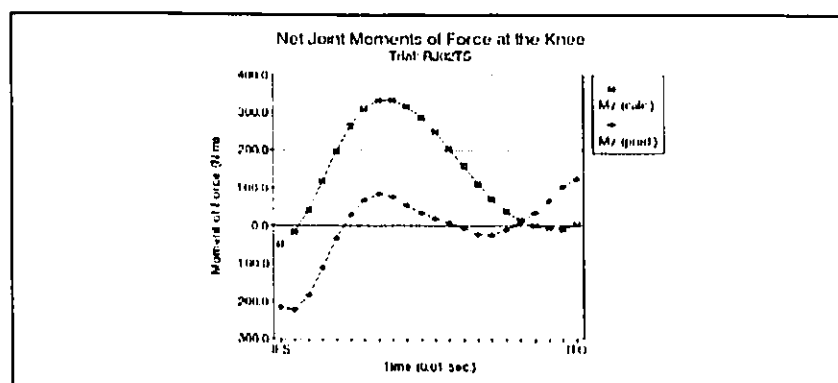
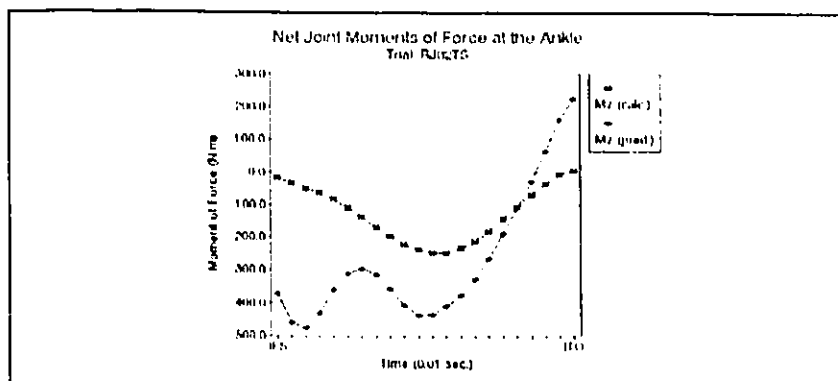


Fig. 5. Examples of calculated and predicted net joint moments of force for the ankle, knee and hip when jogging ($v = 3.8$ m/s).

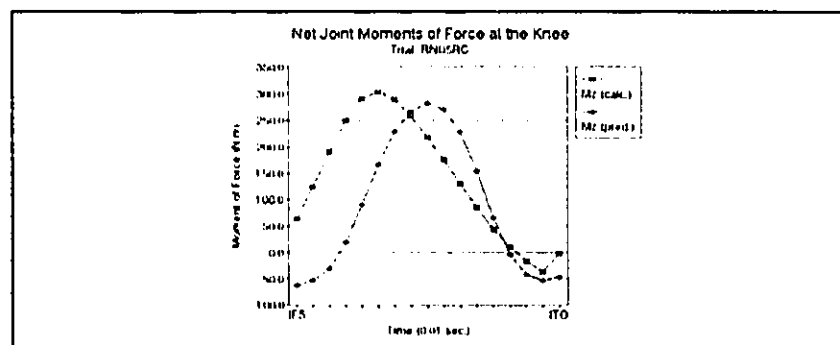
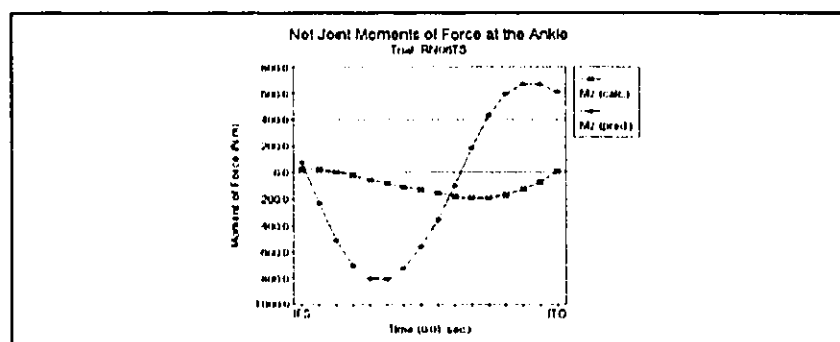
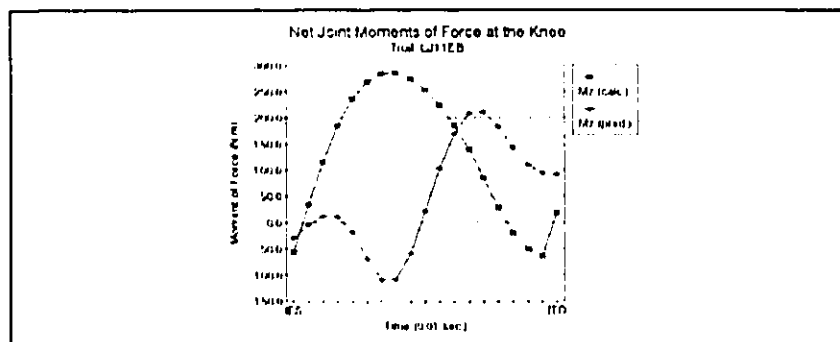


Fig. 6. Examples of the measured and predicted moments of force at the knee and ankle in the long jump takeoff (LJ11EB), sprinting and running (RN06TS and RN05RC). $v = 4.9, 6.1$ and 4.8 m/s respectively.

graph of similar scale, the pattern of the two curves are somewhat similar. This was also the case with the results of a number of the other trials not shown here. The predicted moments at the knee and ankle joints should show distinctly different patterns, as was the case with the criterion. In general, moment results for the hip joint were also not well predicted.

Figure 6 compares the net joint moment results for running, sprinting and the takeoff phase of the long jump. The predicted moment curve at the knee for trial RN05RC, was one of the trials already mentioned that matched the criterion curve extremely well. For this particular subject, the predicted moment curves at the knee were similar to the criterion in all three of the trials (jogging, running and sprinting). The other predicted moment results (ankle and hip joints) did not compare well with the criterion values.

Effect of cutoff frequency and linked segment model on the accuracy of film predicted results

Figures 7, 8 and 9 show the effect of varying the cutoff frequency and the number of limbs used in the linked segment model on the film predicted results. The digital filter cutoff frequency and linked segment modelling parameters were varied for the film prediction method only. The trial depicted in Figures 7, 8 and 9 is for sprinting, and it can be seen that lowering the cutoff frequency to 4 Hz improved the predicted F_x curve, but reduced the accuracy of F_y . For joint moments, the lower cutoff frequency reduced some of the magnitude error for the predicted moment curve at the ankle joint, but had no effect on the overall shape of the curve. Changing the linked segment model by eliminating first the arms and then the swing leg, had little effect on the prediction of F_x , but decreased the accuracy

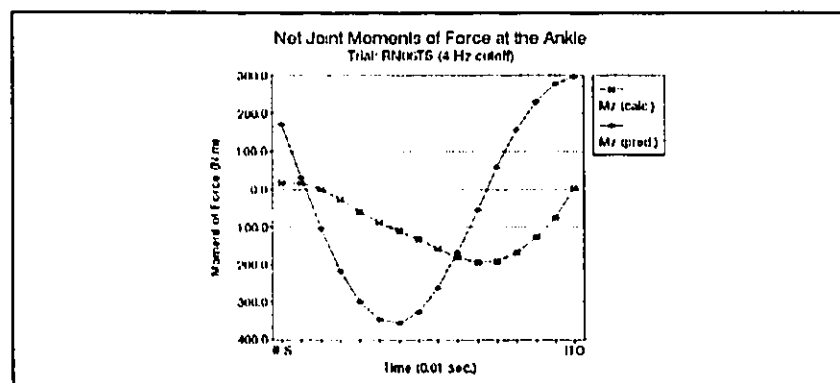
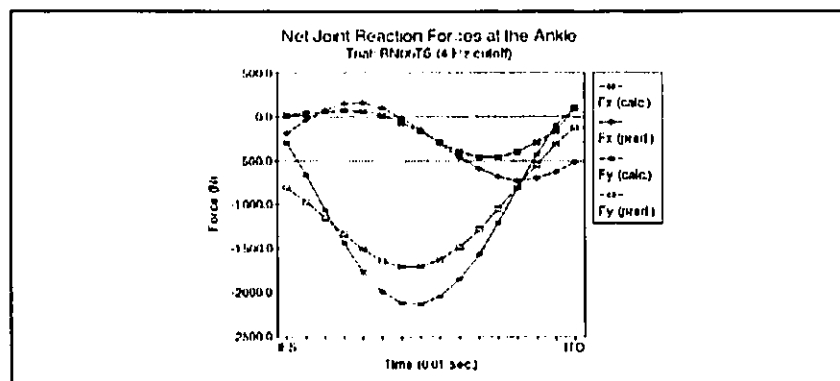
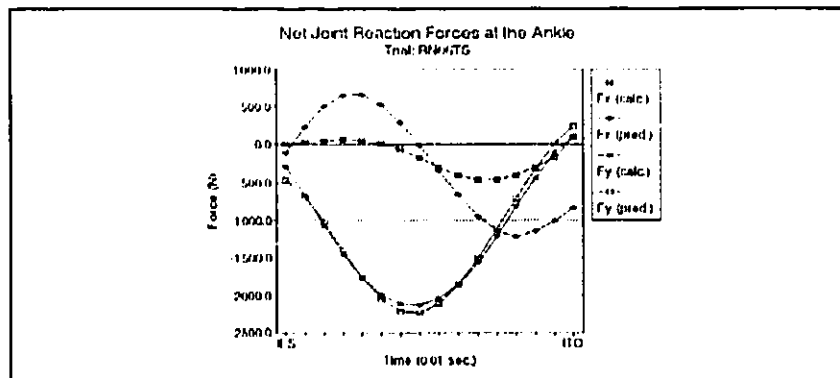


Fig. 7. Examples of varying the filter cutoff frequency on the film predicted force and moment results. Trial RN06TS is for sprinting ($v = 6.1$ m/s).

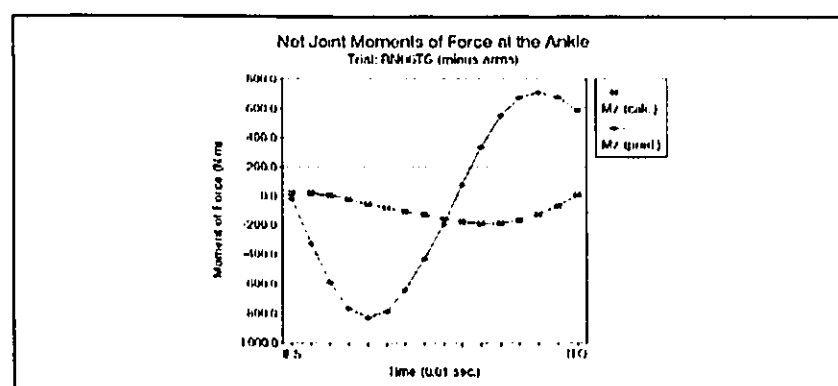
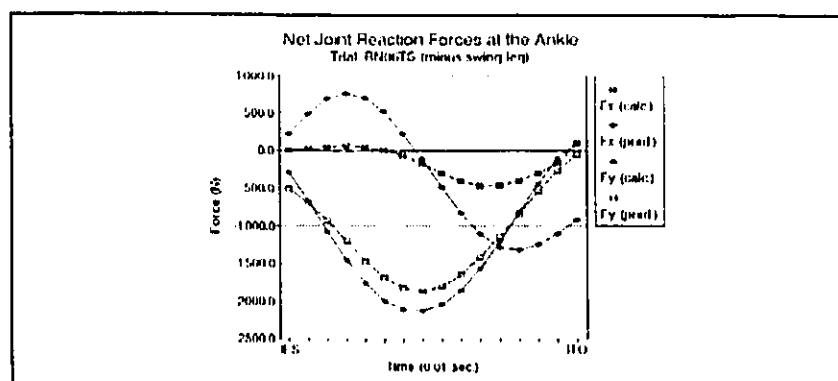
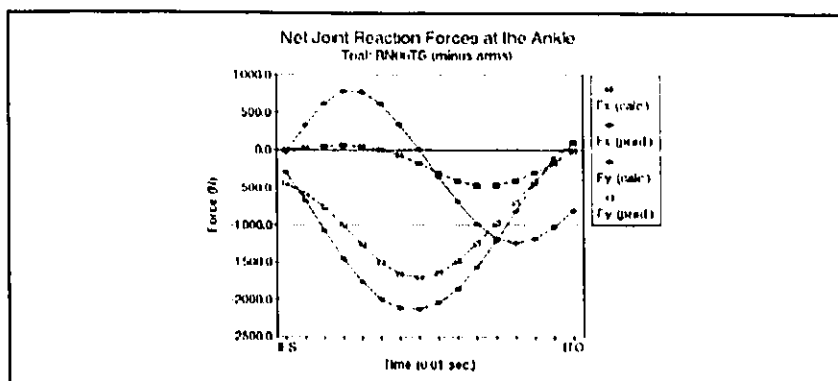


Fig. 8. Examples of the effect of varying the linked segment model on the predicted force and moment results.

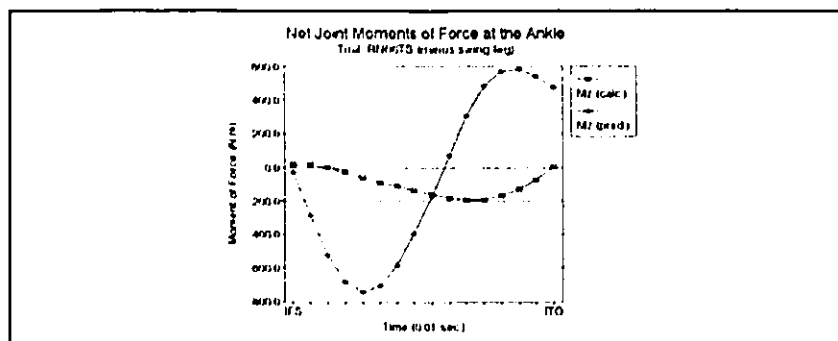


Fig. 9. Predicted and calculated net joint moments of force for the ankle in sprinting, minus the swing leg.

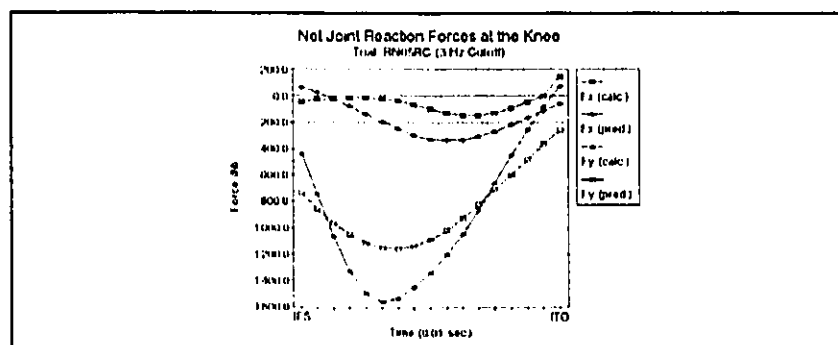
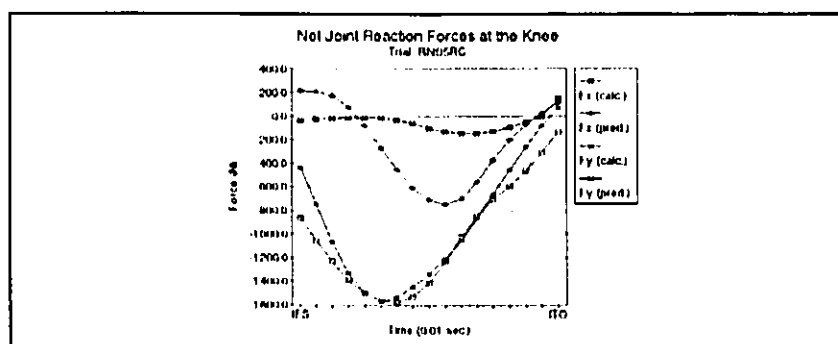


Fig. 10. Effect of lowering the cutoff frequency on the predicted joint reaction forces at the knee in running ($v = 4.8$ m/s).

of predicting F_y . Varying the model had no effect on the predicted joint moment results, other than a slight reduction in the magnitude of the maxima and minima of the predicted joint moment curve when modelled without the swing leg. Figure 10 shows a second example of the effect of a reduced cutoff frequency on the predicted forces at the knee. Once again, the large magnitude error occurring in F_x in the middle of the support phase was reduced, along with a consequent decrease in the accuracy of the prediction of F_y . For this particular subject (RC), the large overestimation of F_x in the middle of the support phase occurred in all of the trials. The other two subjects (TS and EB) did not show the same kind of error at this point in the support phase.

Comparison of statistics

Tables 1 and 2 show the Pearson correlation and RMSE results between the predicted and criterion force curves for F_x and F_y , respectively. Table 3 shows the same parameters for the comparison between the predicted and criterion joint moment of force results.

Examination of the RMSE and Pearson correlation statistics for the joint reaction forces, reveal a number of interesting points. For example, the RMSE was identical in all cases from one joint to another within a trial, for both F_x and F_y (shown in Tables 1 and 2, by the trial RJ02TS). In the majority of cases the RMSE increased as the speed of the motion increased (e.g., from running to sprinting), noticeably more for F_x than F_y . When expressed as a percentage of the maximum range of the criterion curve, the RMSE values for

Table 1 Statistical comparison between measured and predicted Fx curves

Trial	RMSE (N)	r	% of max. range
RJ02TSA	203.30	.745	44.7
RJ02TSK	203.31	.463	66.6
RJ02TSH	203.30	-.066	54.1
LJ11EBK	328.81	.744	50.6
RN06TSA	562.20	.761	100.1
RN06TSA*	249.18	.766	44.4
RN06TSA ⁺	607.48	.777	108.2
RN06TSA [#]	654.90	.799	116.7
RN05TSK	197.74	.669	53.3
RN05RCK	344.44	.688	117.5
RN05RCK [^]	157.83	.659	53.9

* 4 Hz cutoff A = Ankle

+ minus arms K = Knee

minus swing leg H = Hip

[^] 3 Hz cutoff

Fy were generally near or below 10%, whereas the values for Fx were much higher (50% or more). The RMSE results for the joint moments confirmed what the graphs clearly show, that in most cases the predicted moments replicated the criterion moment values very poorly. The RMSE error for joint moments also increased, for most of the trials, as the movements got faster (similar to Fx). This was observed more from running to sprinting, than from jogging to running.

The Pearson correlation statistics confirmed the predicted Fy curves to be very closely

Table 2 Statistical comparison between measured and predicted Fy curves

Trial	RMSE (N)	r	% of max. range
RJ02TSA	198.31	.974	9.5
RJ02TSK	198.31	.973	9.9
RJ02TSH	198.31	.967	11.0
LJ11EBK	340.12	.930	13.9
RN06TSA	91.47	.995	4.3
RN06TSA*	298.77	.975	14.0
RN06TSA ⁺	324.38	.991	15.2
RN06TSA [#]	202.93	.996	9.5
RN05TSK	161.55	.987	7.6
RN05RCK	166.86	.978	10.7
RN05RCK [^]	256.81	.984	16.4

* 4 Hz cutoff A = Ankle

+ minus arms K = Knee

minus swing leg H = Hip

[^] 3 Hz cutoff

correlated with the force platform calculated results. The correlations for the predicted and calculated Fy curves were very similar between joints within a trial, across trials and across the subjects. The correlations between the predicted and calculated Fx curves were quite different from those of Fy. In every one of the 156 joint reaction force graphs that were analyzed, the correlation coefficient for Fx was high at the ankle joint, lower at the knee and very low or negative at the hip (e.g., Trial RN02EB, not shown;

Table 3 Statistical comparison between measured and predicted joint moment curves

Trial	RMSE (N.m)	r	% of max. range
RJ02TSA	221.86	.531	88.4
RJ02TSK	188.18	.415	49.4
RJ02TSH	209.60	.082	64.9
LJ11EBK	210.12	-.501	60.2
RN06TSA	566.36	-.204	266.8
RN06TSA*	227.60	.193	107.2
RN06TSA ⁺	594.83	-.349	280.2
RN06TSA [#]	523.18	-.363	246.4
RN05RCK	119.79	.610	35.2

* 4 Hz cutoff A = Ankle

+ minus arms K = Knee

minus swing leg H = Hip

Ankle: $r = .925$, Knee: $r = .746$, Hip: $r = -.038$. See also Table 1, Trial RJ02TS).

Generally speaking, the correlation coefficients for Fx were lower than for Fy and showed variability between joints within a trial and, to a lesser extent, across trials (whereas Fy showed little or no variability). The correlation results for net joint moments indicate that the predicted curves are not consistently correlated with the criterion results in any way. A lot of variability exists among the r values calculated for the joint moments. A few trials were correlated well both graphically and statistically (e.g., Trial RN05RC, $r = .610$; see Table 3 and Fig. 6.), but most were poorly or even negatively correlated.

Both the statistics and graphs indicate that the predicted Fx reaction forces parallel the measured forces well with respect to the pattern of the motion at the knee and ankle joints (shown by the degree to which the two curves are correlated), but do not parallel them well in terms of the magnitude error (quantified using the RMSE) at either of the three joints. This error, which occurred in almost all of the trials at one point or another in the support phase, could be reduced by lowering the cutoff frequency of the filter, as shown by the force graphs (see Fig.'s 7 and 10) and the RMSE results presented in Table 1. The RMSE results also show to what extent the accuracy of Fy (pred.) was reduced when using the lower cutoff frequency.

DISCUSSION

Comparison of measured and predicted force curves

The purpose of this study was to predict the net joint reaction forces and moments for a number of different motions using displacement data, and to validate these estimates by comparing them to the force platform determined results. The criterion vertical and horizontal joint reaction force histories were similar across all trials of a particular motion. The findings indicate that accurate vertical net joint reaction forces (Fy) can be predicted for the movements of jogging, running, sprinting, the long jump takeoff and a front somersault into a forward roll, using the methods employed in this investigation. The accuracy in predicting Fy proved to be independent of subject, trial, movement or joint. The RMSE and

Pearson correlation statistics quantify what the graphs of the predicted and criterion curves indicate visually. In one particular trial (LJ11EB) the RMSE for F_y was inflated due to the fact that the predicted curve was slightly out of phase with the criterion. This may have occurred as a result of modelling the two lower limbs as a single support limb in the method used to calculate the predicted results. In all of the trials, the predicted and calculated curves for F_y were correlated at a level higher than $r = .900$, and with a few exceptions the RMSE was less than or equal to 10% of the range in magnitude for the force platform calculated curve.

Net joint reaction forces were not predicted as well for F_x , and generally showed a substantial amount of error in magnitude. The magnitude error seemed to increase proportionally with the speed of the movement (i.e., higher for sprinting than jogging; compare Trials RJ02TS and RN06TS). More importantly, the correct horizontal force pattern could be predicted from displacement data for both the knee and ankle joints.

The correlation results revealed that the predicted F_x was not correlated as well for the hip as it was for the ankle or knee. The Pearson correlation statistic is very sensitive and can be affected by error in one small part of the curve, as evidenced by Trial RJ02TS. Visual examination of the graphs reveals no major difference in the predicted and criterion curves of the three joints, although the correlation is in fact negative for the curves at the hip. This is due, presumably, to the error occurring in the first 5/100ths of a second following heel-strike. The force platform calculated history for F_x is similar for the knee and ankle joints, but slightly different at the hip. The predicted F_x pattern was the same for all three of the support leg joints in all of the trials. It is apparent that the predicted method

is not sensitive to this slight difference in F_x at the hip. This difference is not immediately noticeable but can be seen when one examines the graphs of the predicted and calculated force curves at each joint, a second time. The error in predicting F_x at the hip may have been caused by inaccuracies in the estimation of the position of the hip marker on the far side of the body, or in attempting to maintain a stable trunk segment length between the shoulder and hip markers to minimize the effect of trunk segment kinematic error.

One of the subjects (RC) showed predicted F_x magnitude error occurring in the middle of the support phase (see Fig. 10), in contrast to error at the beginning and end of the support phase observed in the trials of the other two subjects. This difference between subjects may have been due to the fact that subject RC was female and required different body segment parameters from the other two subjects, who were male. In addition, this particular subject ran differently, with the ball of the foot making the initial contact with the ground, not the heel. An iliac crest marker was also not used for this subject because of clothing restrictions (the subject wore a sleeveless shirt), and this may have had some effect on the accuracy of trying to maintain a consistent trunk segment length when digitizing, which was facilitated by the inclusion of this marker with the other two subjects.

Experimentation with the cutoff frequency used to filter the displacement data yielded some interesting results. Clearly, the large magnitude errors seen primarily in the faster movements for F_x could be reduced by decreasing the cutoff frequency from 6 Hz, to 4 Hz or even 3 Hz, depending on the size of the error. A number of different cutoff frequencies ranging from 2 to 15 Hz were examined. Cutoff frequencies of 6 Hz were sufficient for F_x in the jogging trials, but needed to be lowered to obtain equivalent results in the running and,

especially, sprinting trials. While the prediction of F_x was improved by lowering the cutoff frequency, F_y decreased in its predictive ability. These trends are indicated by both the RMSE and joint reaction force graphs (see Fig.'s 7 and 10). The different effect produced by the lower cutoff frequency on F_x and F_y , might be due to differences in the marker position signals for the x and y directions. In examining the range of displacement travel for all of the markers in both the x and y directions, the amount of displacement covered in x is much greater than that in y . While all the markers move a great deal in the x direction, most (including the markers of the foot) move only minimally in y by comparison. It is also possible that the movement of soft tissue relative to the bony structures referred to by Bobbert et al. (1991), Sanders et al. (1991) and others, causes more error in x than in y . This is not what one would expect, given that most of the impact force peak occurs in the vertical ground reaction force component, and that high frequency noise should be filtered out by the low-pass filter. It is also possible that the improvement in the predicted F_x curve for the faster motions, occurs as a result of oversmoothing or straightening out of the curve when the cutoff frequency was lowered. The explanation of improvement due to oversmoothing seems unlikely, considering the change in the accuracy of F_x with respect to the speed of the movement and the consistent correlation of the predicted F_x curve, both of which exist regardless of whether the error in magnitude is at an acceptable level or not.

Varying the components of the linked segment model used to predict the joint kinetics, indicated that eliminating either the arms or swing leg segments had little effect on the prediction of F_x . Eliminating these segments did have an effect on the ability of the model to predict F_y , in that the accuracy compared to the criterion curve was reduced. This

is logical in light of the fact that eliminating one or two limbs subtracts a portion of the body's total mass from the determination of F_y for each joint. Interestingly, eliminating both arms resulted in a greater loss of accuracy for F_y than the elimination of the swing leg, which has a higher mass proportion than the combined proportional mass for both of the arms. Clearly, the total mass of the body (preceding the joint in question) is needed to predict F_y as accurately as possible from positional data. This supports the notion that the gravitational acceleration is the key component in the prediction of F_y .

Differences in the ability of the displacement data method to accurately predict vertical and horizontal joint reaction forces may be due, in part, to gravity and differences in the magnitude of the inertial accelerations occurring in each direction. The force due to gravity is an ever present and somewhat dominating term in F_y , especially for the support leg joints when most of the total mass of the body has already been accounted for. Gravitational acceleration in F_y may be masking inertial acceleration error which is otherwise undisturbed in F_x , or it may be that the magnitude of the inertial F_y acceleration is small by comparison, causing it to be more or less insignificant. It is possible that the difference in the ability to predict F_x and F_y shows the actual limitation of the ability of displacement data methods to calculate inertial body segment accelerations, especially in the horizontal direction. This hypothesis is consistent with the different effect produced on F_x and F_y by changing the cutoff frequency of the filter. It appears that the digital filtering of vertical and horizontal segment accelerations should be treated differently to obtain the most accurate results.

Comparison of measured and predicted moment curves

The net joint moments of force about the transverse axis (M_z) could not be consistently predicted using the particular method employed in this study. The force platform calculated moment of force histories were identical in pattern across all subjects, movements and trials and were therefore considered to be a suitable and valid criterion. The predicted moments were, in most cases, not sensitive to the existing joint differences. In some trials, the patterns of the predicted joint moments were similar for all three joints, when they should have been different. A number of trials showed accurately predicted joint moment patterns for one joint, but no resemblance to the criterion values for the other two joints. In addition, there seemed to be no relationship between the ability to predict joint reaction forces and the ability to predict net joint moments. In some cases the joint moment pattern was predicted well at a certain joint, while the horizontal joint reaction forces were substantially overestimated. In other cases, F_x was predicted very accurately while the joint moments showed a large amount of correlation and RMS error.

The statistical results, like the graphs, showed no consistency for the RMSE or Pearson correlation comparisons. RMSE values were usually higher than 50% of the maximum range of the directly measured moment curve, and the Pearson correlation coefficients were low or negative for many of the trials. The highest correlation for a predicted joint moment curve was $r = .9649$, with a RMSE of 34.8% of the maximum range in M_z (Trial FR12EB, not shown). This was typical of the results for the predicted joint moment curves; if they happened to accurately depict the calculated moment curves, they

were in many cases either slightly out of phase or overestimated the criterion, or both. The predicted moments estimated the pattern of the criterion curves better than the magnitude; i.e., there were a number of acceptable Pearson correlations, but no acceptable (none < 10%) RMSE values.

Changing the components of the segment model had almost no effect on the predicted joint moment curves other than a very slight change in the RMSE, and no effect on the configuration of the curve (see Table 3). Lowering the filter cutoff frequency did produce some improvement in the RMSE, but had no effect on improving the correlation between the predicted and calculated curves.

It is difficult to speculate on the degree to which the method employed in this study is successful or unsuccessful in predicting net joint moments of force from displacement data. For some trials at certain joints the method seemed to work, whereas in other trials and at other joints it did not. It is difficult to say that a method is completely ineffective, when it is in fact minimally successful a certain percentage of the time. Based on the results of this study, it must be concluded that it is not possible to predict net joint moments of force from displacement data using this method with any degree of certainty. This does not mean that it cannot be done, but that it is not accurate or reliable enough to be of any practical use at the present time. With an easy and accurate procedure to estimate the centre of pressure of the ground reaction force vector directly from the displacement data, it is possible that accurate net joint moments of force could be predicted using the method applied in this study. The prediction of net joint moments from reaction force and inertial moment of force information only, is clearly not acceptable or sufficient.

CONCLUSIONS

This study has demonstrated the ability to accurately predict the vertical net joint reaction forces at all three support limb joints using the inverse dynamics approach and a total body, linked segment rigid body model. It has also demonstrated the ability to accurately predict from displacement data the pattern, and in some cases the magnitude, of the horizontal net joint reaction forces for the knee and ankle joints in a number of different, relatively fast, single support, sagittal plane human locomotor movements. No consistent success was achieved in predicting the net joint moments of force using the method employed in this study. More work needs to be done if net joint moments are to be predicted accurately from displacement data using indirect methods.

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APPENDIX

REVIEW OF LITERATURE

In this chapter a review of the prominent research relevant to the understanding of the major principles involved in determining the joint kinetics of human motion is presented. In addition, the methods and results of previous research pertaining specifically to the derivation of ground forces and joint kinetics from displacement data are reviewed.

Force platform design and applications

A force platform is a fundamental research tool used in the biomechanical analysis of human movement and is particularly useful in studies involving locomotor movements. In order to be able to quantitatively describe human motion, three different types of essential data are necessary. The first involves the amount of displacement incurred by the total body and/or its individual body segments. The second concerns the specific parameters of those body segments (the centre of mass, moment of inertia, radius of gyration, etc.), while the third pertains to the forces that are imposed on (or by) the total body or body segments. Most often the displacement data are obtained from cinematographic techniques, the body segment parameters from one of several anthropometric models, and the force information from direct measurements using some type of force transducer (Ramey, 1975).

In a locomotor activity such as walking, the purpose of a force platform is to measure the magnitude and direction of the forces exerted by the ground against the foot (i.e., the ground reaction forces). These forces are in fact complex pressure distributions which are

applied across different parts of the foot, and which vary in time according to a number of factors including the applied forces of the contact, the frictional characteristics of the surfaces, the shape of the foot and the position and rigidity of the platform itself (Robertson, 1984). As Ramey (1975) has stated, even though there have been "numerous design concepts used in the development of force plates, they all reflect the fact that the force plate is basically a weighing system that responds to changes in the displacement of a sensing element. The sensing element is chosen such that the force acting on the element is directly proportional to the displacement of the element". In most situations this displacement is quite small and requires either some form of amplification or the use of recording devices that are extremely sensitive to these small displacement changes. Force transducing devices can usually be classified into one of four different sensing element categories: (1) mechanical springs and pointers, (2) electric resistance strain gauges, (3) linear variable differential transformers (LVDTs), and (4) piezoelectric crystal devices (Ramey, 1975).

Two of the earliest force platform designs were those of Amar (1916) and Elftman (1938). Ramey (1975) describes Amar's "Trottoir Dynamographique" as an instrument equipped with a mechanical measuring system which operated on the principle that a load induced spring deflection would manipulate a number of associated indicators. These indicators measured the vertical, medio-lateral and backward (no forward) force components. Amar used this instrument in his investigations of the locomotion of amputees with artificial limbs, his purpose being to use these results to suggest any necessary design improvements. Elftman (1938) used a similar design concept in attempting to measure the magnitude and position of the resultant force during the support phase in walking. Like Amar's design,

Elftman used strictly a mechanical system which utilized linear springs as the sensing element and mechanical levers as the recording device. These indicating levers were photographed with a high-speed cinecamera to record the changing force magnitudes throughout the entire support phase of the foot. A combination of two platforms, situated one on top of the other and separated by ball bearings, were used to measure the components of force in the vertical and both horizontal directions (i.e. fore-aft and medio-lateral).

In 1952, Cunningham and Brown designed a six-component dynamometer which used electrical resistance strain gauges connected in bridge circuits as the sensing element. The outputs from this particular force platform design included the three components of the resultant force, in addition to the moments about the three perpendicular axes through the centre of the plate (Paul, 1976). The electrical resistance strain gauge, which is fundamentally a short length of wire or a thin foil, is typically glued to a metal sensor which strains under the influence of an imposed load. As the sensor is deformed, so too is the strain gauge. The material of which the strain gauge is made changes electrical resistance in proportion to the deformation, and by the use of proper electrical circuitry this change in resistance can be used as a measure of the force imposed upon the sensor (Ramey, 1975). Two examples of the many force platform designs which have employed electrical resistance strain gauges, are that of Whitney (1958) and Ramey (1970).

In 1958, Greene and Morris developed a force platform which used a linear variable differential transformer (LVDT) as the sensing element. An LVDT is a device which, through the use of a magnetic core surrounded by three electrical coils, is able to easily convert mechanical displacement into a proportional electrical signal. Two other researchers

who designed force platforms based on the LVDT concept were Whetsel (1964) and Hearn and Konz, who in 1968 introduced an improved design which used a total of nine LVDTs to measure the torques produced about three mutually perpendicular axes, in addition to the standard three orthogonal force vector components (Ramey, 1975).

In 1954, Lauru was the first to implement piezoelectric quartz crystals as the sensing element in a force platform design. What is known as the "piezoelectric effect" occurs when certain solid materials respond to a change in shape by producing an electric charge. Quartz is one such material which exhibits this effect, and "in force plate applications the quartz crystal is usually placed at the corners of the platform and responds to the deformation caused by the forces in one direction only, by producing a voltage proportional to the force. After proper amplification the output voltage is displayed on readout devices that include oscilloscopes, oscillographs, and magnetic tape", or the signal can be fed directly into a computer equipped with analog to digital (A/D) conversion facilities where digital filtering and other signal processing operations can be performed (Ramey, 1975). Lauru's force plate was referred to as the "Effort Detector" and measured the components of force in the three standard directions, but experienced problems due to the fact that the capacitive nature of the quartz element did not allow the crystals to respond well to static forces. Since the time of Lauru's initial effort with piezoelectric crystals, more recent applications of quartz elements have benefitted from improvements in signal conditioning. Better charge amplifiers, for example, have greatly improved the static measuring capability of these types of force plates. One of the main benefits of piezoelectric force platforms is that the stiffness of the quartz crystals gives these systems very good dynamic response characteristics (Ramey, 1975).

In more recent years, commercially marketed piezoelectric force platform systems have become available that are specifically designed for biomechanics research. An example of a system currently on the market is that manufactured by Kistler Instrumente AG of Winterthur, Switzerland. The Kistler force platform, reputed to be the best design currently available, is a six quantity dynamometer which, according to the manufacturers, will yield force measurements with errors not greater than 2% (Paul, 1976; Bobbert and Schamhardt, 1990). Some of the important features to be considered in any force platform design, and which contribute to the success of the Kistler type, are the linearity of the sensors, the ability of the platform to be able to sense forces in one direction independent of the forces acting in any of the other directions (i.e., no cross-talk between axes), the dynamic characteristics of the system and the ease and accuracy of the calibration procedures (Ramey, 1975).

The invention of devices which can quantitatively measure the forces associated with certain types of human movement, has allowed a variety of investigations to be performed. Force platforms have been used to quantify the forces in gait and human locomotion studies, to perform time-motion studies for industrial activities, and to provide information for the assessment and improvement of prosthetic devices. They have also been used as essential research tools in human rehabilitation studies, physical fitness and motor skills testing, and in a variety of military applications and sports skills analysis programs (Ramey, 1975). Studies by Rehman (1947), Peizer (1968) and Peizer and Wright (1969) have used force plates to characterize gait patterns for both normal and pathological conditions in both the young and the elderly.

Force platforms have also been used in many studies pertaining to sports science research. Some examples include Payne et al. (1968), who used a force plate and cinematography to examine the mechanics involved in the sprint start, vertical jumping, constant speed running, hurdling, shot putting and weight lifting, Kuhlow (1973) who studied the dynamic variables involved in both the flop and the straddle styles of high jumping, Barlow (1973) who studied the pole vault using a force plate to measure the forces associated with vaulting, and Robertson (1987) who analyzed the contributions made by the leg muscle groups to the external mechanical work in both the vertical and standing broad jumps (Ramey, 1975; Robertson, 1987). As evidenced above the force platform has been, and will continue to be, a prominent and multi-faceted tool in the diverse field of biomechanics research.

Body segment parameters

To conduct a detailed quantitative analysis of any human motion, information concerning the inertial characteristics of the total body and/or its individual segments are required. When applying an n-link rigid body model to the movement analysis in question, the inertial characteristics of all the specified segments in the model must be known. A method referred to as the *indirect (or segmental) approach* is used, in which the various segments of the body are considered separately in determining these characteristics. In this approach the segmental values can also be used in summation, if necessary, to compute whole body values such as the total body centre of gravity. The term inertial characteristics,

or *body segment parameters*, refers to the quantities of mass, centre of mass location, moment of inertia and radius of gyration to define a particular body segment. Investigations which have sought to determine these body segment parameters can be classified into one of three different categories: (1) cadaver studies, (2) mathematical models and (3) other methods.

Research into the inertial characteristics of individual segments of the human body began with the dissection of cadavers. Harless (1860) was the first to perform a cadaver study for these purposes, in which he dissected the cadavers of two executed criminals (Hay, 1973). In 1889, Braune and Fischer dissected four male cadavers and provided data on segment weights and centre of gravity locations that have been used extensively in many early studies (Hay, 1973). Fischer performed another cadaver study in 1906, which used the compound pendulum method to determine the moment of inertia for selected segments (Hay, 1974). Using the data from this study, along with the segment mass and centre of gravity locations previously determined by himself and Braune, Fischer was able to compute the moment of inertia and radius of gyration of each segment relative to a transverse axis through its centre of gravity. The data of Fischer (1906) has been used in many early investigations into the mechanics of human motion, two examples of which are the work of Elftman (1939a) and Bresler and Frankel (1950). Fischer's work is worthy of notation, in that it represented the most widely quoted source of data for a period of almost 50 years until the next major cadaver study was conducted by Dempster in 1955 (Hay, 1973).

Dempster's (1955) cadaver study involved eight, white male, "medium build" (Dempster's term) cadavers ranging in age from 52-83 years, with a mean age of 68.5.

Dempster's results included values for the moment of inertia, the mass, and the centre of gravity location of a number of different trunk and limb body segments. Dempster's data superseded that of Braune and Fischer, and has been used in numerous studies and cited frequently in the literature, primarily because of minor procedural improvements in dissection and the greater number of subjects used in his investigation.

After Dempster's study, Barter (1957) attempted to overcome the inherent limitations imposed by the small sample sizes in the studies of Braune and Fischer (1889), Fischer (1906) and Dempster (1955) by combining their results for the different segment weights together and treating them statistically by means of a regression analysis. Hay (1973) mentions that Barter was aware that such an approach ignored differences in the dismembering techniques used in the three studies, but considered that these differences were probably not significant when the magnitude of the errors introduced by other factors were considered (especially sample size). Barter's contribution to improving the accuracy of the existing anthropometric literature was specific to only one parameter, that being the estimation of the mass of a body segment.

Another cadaver study of significance was that of Clauser et al. (1969), in which 14 male cadavers, preserved and carefully selected to closely approximate the wide range of physical body sizes in normal populations, were dissected. Extensive care was taken to eliminate fluid loss and tissue loss during dismemberment, and once the required segment parameters were determined regression equations were derived for the prediction of segment weight, volume and centre of gravity location through the use of a step-wise multiple regression analysis procedure (Hay, 1973). Clauser et al. concluded their investigation by

stating that they believed the predictive equations developed in their study, provided a better estimate of the parameters analyzed than had previously been available. To this they added the comment that "they should not, however, be considered as other than good first approximations until they can be adequately validated on live populations" (Clauser et al., 1969). It is interesting to note that this study by Clauser et al. did not provide any data concerning values for the moment of inertia of the various body segments.

In contrast to the use of cadavers, a number of investigators have employed a completely different approach in attempting to determine the inertial parameters of body segments. What is referred to as the *mathematical modelling approach* (or method), represents each body segment as a regular geometric solid of known mass in order to compute the required segment parameter values (Hay, 1974). Amar (1920) was one of the first to utilize a mathematical modelling approach, when he considered the trunk to be a cylinder and the limbs to be frusta of cones in computing the segmental moments of inertia for a 65 kg adult male (Hay, 1974).

Much later, Whitsett (1963) developed a 14 segment model of the human body for the purpose of predicting man's mechanical behaviour in some selected conditions associated with weightlessness. Principal moments of inertia were computed using the standard equations for the corresponding geometric solid (Hay, 1974). Shortly thereafter, Hanavan (1964) designed a 15 segment model almost identical to that of Whitsett, in which "the dimensions and properties of the segments were calculated using a total of 25 anthropometric measurements taken from an individual subject" (Hay, 1974). Barter's regression equations were used to predict the mass of each of the model's segments. Hanavan attempted to

validate his model by comparing his data to that of Santschi et al. (1963), who had computed the principal moments of inertia of the living human body in eight different positions.

Hanavan reported that one half of his predicted values for segment moments of inertia about the frontal and transverse axes, fell within 10 percent of the experimentally obtained values determined by Santschi et al. (1963). In addition to this, Hanavan also compared his results to those obtained by Dempster (1955), and concluded that "the very small deviation between the model and the experimental results indicates that the shape and size of these segments approximate the body segment very well" (Hanavan, 1964 in Hay, 1973).

With respect to the practical use of mathematical models, Hay (1973) states that one should be aware that assumptions must be made about one or more of: (1) the segment densities, (2) the segment centre of gravity locations, and (3) the geometry of the segments themselves. The mathematical models of Whitsett (1963) and Hanavan (1964) are worthy of mention in this section, in that they provide values for the segmental moments of inertia about all three principal axes, whereas other studies such as Dempster's (1955), give segmental moment of inertia values relative to the transverse axis only. In comparing the models of Hanavan and Whitsett, Hay (1974) considers Hanavan's to be more appropriate because it is based on the physical dimensions of individual subjects rather than on mean values.

Another type of mathematical model, originally conceived by Weinbach (1938), involves a technique which depicts the body as being made up of a number of identical elliptical zones. In 1976, Jensen incorporated elliptical zones into the development of a body segment parameter model, the purpose of which was to: (1) "give the segmental and the

whole body volume, mass, buoyancy, mass centre, and moment of inertia, (2) develop apparatus to determine dimensions and masses and (3) compare the calculated parameters with parameters obtained from the model developed by Hanavan in 1964" (Jensen, 1976). Part of the rationale for the development of Jensen's model using the elliptical zone method, was that segment shapes "cannot be duplicated by a geometric solid of revolution", this being a major limitation of Hanavan's model (Hanavan, 1964 in Jensen, 1976). Jensen's (1986) method involved using a 16 segment human body model made up of elliptical zones 2 cm wide. Although values for segment density had to be obtained from the literature (Clauser et al., 1969), one of the main advantages of Jensen's method is that it calculates parameter values that are directly specific to the subject being measured. The mathematical models of Hanavan (1964) and Whitsett (1963) are also specific to live subjects, more so than that of data obtained from cadaver studies, but depend more on input data from the existing literature than does Jensen's model. For example, although Whitsett's (1963) model measured the dimensions of the segments directly from the subjects, the masses of the limb segments were calculated using Barter's regression equations and the segment centres of gravity obtained using Dempster's (1955) cadaver data (Hay, 1973).

Along with cadaver studies and mathematical modelling techniques, a number of other methods have been developed to determine segment parameter values for the human body. One of the best examples is the *gamma-scanner (or radioisotope) method*, which determines segment density, mass, centre of mass and moment of inertia by recording the degree of weakening of a gamma radiation beam as it is passed through the tissue of the body (Zatsiorsky and Seluyanov, 1983). In a study which used 100 live, male subjects, Zatsiorsky

and Seluyanov (1983) employed a 16 segment model to determine values for the above mentioned segment parameters. Based on their results they developed regression equations to predict these parameters, using primarily the independent variables of total body height and weight as the predictors.

With all of the different sources of data that are available, one can sometimes be left wondering which set of body segment parameter data is the most appropriate for use in human movement analysis experiments. It can be a difficult decision, primarily because of the number of different assumptions made in each of the wide variety of methods used to predict this information. Hay (1973) states that as a result of "the absence of any clear-cut guidelines concerning the validity of these assumptions, it would seem logical to select the data which has been gathered on subjects who most closely approximate the subjects under investigation in age, height, weight, sex, race, physique and health status". All of these are factors certain to have some influence on the accuracy and applicability of the chosen body segment parameter data to the subjects in question.

Inverse dynamics

Within the broad discipline of mechanics, dynamics is defined as that section which deals with the motion of bodies under the action of forces (the other section being statics, which deals with the action of forces on bodies at rest). Dynamics itself can be further subdivided into two distinct parts or levels. The first, *kinematics*, is the study of motion

without reference to the forces which cause that motion, and the second, *kinetics*, relates the action of forces on bodies to their resulting motions (Meriam and Kraige, 1987).

In the study of human motion it is usually the more comprehensive level, kinetics, that is the subject of interest as it is often the most intriguing and least understood aspect of the motion being studied. When conducting a dynamic analysis at the kinetic level, two distinct approaches are possible. Meriam and Kraige (1987) describe the first as a situation in which the forces are specified and the resulting motion is to be determined. This is generally referred to as the *direct* approach, and in the context of human motion studies it would involve a situation where the required forces are measured directly using some type of dynamometric device, and are then used to solve for the system kinematics. The second approach, usually referred to as the *inverse* approach, involves a situation where the acceleration is either specified or can be determined directly from known kinematic conditions. The corresponding forces and moments of force which act on the body whose motion is specified, can then be determined using the equations derived from Newton's second law of motion (Meriam and Kraige, 1987).

Utilization of the inverse approach is quite common in human motion studies involving locomotor activities, in which joint reaction forces and net joint moments of force are most often the quantities to be determined. Winter (1990) has stated that "if we have a full kinematic description, accurate anthropometric measures, and the external forces, we can calculate the joint reaction forces and muscle moments. This prediction is called an *inverse solution* and is a very powerful tool in gaining insight into the net summation of all muscle activity at each joint."

Elftman (1939a) was the first to quantitatively determine the reaction forces and moments of force of the individual joints in a work-energy analysis of walking, in which he implemented the inverse dynamics approach. Elftman used cinematography to measure kinematic variables, Fischer's (1906) data for determining anthropometric properties, and a force platform to measure the components of the ground reaction force needed to solve for the above joint forces and joint moments of force. Bresler and Frankel (1950) also utilized the inverse dynamics approach in an investigation of the forces and moments in the leg during level walking. Their study was significant in that it was the first detailed and comprehensive analysis of the mechanical functions of the legs in walking, determining all three orthogonal components of force along with moments of force in two directions for each joint of the leg. Moments about the z-axis (vertical) were omitted, due to the fact that angular accelerations about this axis were very small and could not be computed without the aid of measurements concerning the twisting of the segments (i.e. rotational data).

In the years following the study of Bresler and Frankel (1950), numerous investigators have made use of the inverse dynamics approach in their research, one example being Dillman (1970) who studied the kinetics of the recovery leg in sprint running. Reliable sources include Miller and Nelson (1973) and more recently Winter (1990), who have both provided accurate and detailed descriptions of the theory involved in applying the inverse dynamics process. In addition, a recent paper by Sanders et al. (1991) has investigated the accuracy of the inverse dynamics approach in deriving the ground reaction forces in an impact oriented activity.

Derivation of ground forces and joint kinetics from displacement data

In determining the joint kinetics associated with human locomotion, the common approach involves measuring the components and position of the ground reaction force (GRF) vector using a force plate or some other type of force transducer. This information is then combined with the kinematics and inertial characteristics of the various body segments, and through the use of the *inverse dynamics* approach the reaction forces and net moments of force at the joints can be computed. For a variety of different reasons, the use of a force platform to measure the ground reaction forces is not always possible or practical. If information concerning the ground reactions and/or the joint forces and moments is still required, it must in these particular situations be derived using kinematics that have been either measured directly or obtained from the double differentiation of displacement data.

One of the first attempts at predicting ground reaction forces from gait data was made by Thornton-Trump and Daher in 1975. Using gait data collected from a number of different sources they designed a computer model to predict ground reaction forces and joint moments, with the intention of using this information as the basis for the design of prosthetic polycentric knee joint mechanisms. The locomotion studies of Murray (1967) along with anthropometric data from Dempster (1955), were combined to synthesize a three-dimensional data set, which was then used by the computer model to predict lower limb joint moments and ground reaction forces. The predicted ground reaction forces were compared to the force plate data of Cunningham (1950), and the predicted moments about a transverse axis at the hip were compared to the hip moments determined by Bresler and Frankel (1950). With

respect to the general pattern, the predicted vertical ground reaction forces matched well with those reported by Cunningham (1950), but showed large variances with respect to magnitude (up to 35% of body weight). This error might possibly be explained if different walking speeds had been used in the studies of Murray (1967) and Cunningham (1950). The moments predicted at the hip show little useful comparison to those reported by Bresler and Frankel (1950), which is not surprising considering the potential error involved in comparing different subjects under different testing conditions. This study is unable to make a satisfactory contribution to the validation of predicted ground reaction forces and joint moments, due to the fact that intertrial and intersubject differences in gait patterns along with possible variations in the walking speed used in the different studies, make useful comparison virtually impossible.

In the same year, Smith (1975) compared computed forces from cinefilm data with measured forces provided by a force platform, and then used the calculated joint forces to estimate the internal forces in the individual muscles of the legs occurring in a drop landing. A Kistler force platform was used to measure the force of the ground on the foot, anthropometric data was taken from Whitsett (1963), and the motion of the drop landing was recorded using a Bolex spring-driven camera operating at 68.5 frames per second (fps). Due to errors "incurred in displacement measurement from the cinefilm, in smoothing and differentiation procedures, and in the use of standard masses for the body segments", Smith sought to implement a check on the accuracy of the calculated inertia forces (Smith, 1975). This check was made by comparing the "computed total thrust at the feet with the measured total thrust as indicated by the force platform" (Smith, 1975). Smith (1975) reported that

"good agreement between the laboriously calculated force curve and that recorded by the force platform system provides reasonable confirmation of the analytical procedures". He also mentioned that "satisfactory correspondence between the calculated and measured force curves was achieved for the drop landing and also for the takeoff action of a standing long jump" (Smith, 1975). In addition to not specifying any definition for "satisfactory correspondence", Smith did not present any data related to the comparison of force platform and cinefilm derived force-time histories. Apparently this is due to the fact that the primary focus of the paper was to estimate the forces occurring in the individual muscles of the legs. It is unfortunate that the required data was not presented in order to substantiate the statements made concerning the apparent "good agreement" between the two types of GRF data. In light of the fact that there is considerable difficulty associated with deriving an accurate representation of the vertical force component from displacement data at the moment of impact, one would expect that Smith's (1975) derived ground forces in the drop landing compared less favourably than those obtained in the takeoff of the standing long jump.

A few years later, Vaughan (1980) was able to show that accurate contact forces could be obtained using a cinematographic procedure. The magnitude of the vertical forces applied to a gymnast by a trampoline bed was investigated along with the resultant moments exerted at the hip and knee joints for different trampoline stunts. Cinematography was used because of the obvious difficulty associated with attempting to instrument a trampoline for the purposes of force data collection. Movements on the trampoline were filmed at 64 fps, with segmental mass and centre of mass data taken from Dempster (1955), and moment of inertia data taken from Whitsett (1963). Raw and time differentiated displacement data was

smoothed using standard cubic spline techniques. The impulse-momentum relationship was used to estimate the accuracy of the acceleration data obtained from film. The area under the acceleration curve was multiplied by the gymnast's mass to determine the impulse. This was validated by calculating the change in momentum, obtained by taking the difference in the gymnast's velocity before and after contact with the bed and multiplying by the body mass. These velocities were calculated by substituting the heights dropped and raised into the equations of uniformly accelerated motion (Vaughan, 1980). Vaughan reported that in the majority of cases the agreement between impulses was extremely high (between 1% and 2% differences). Given that mass is a constant, the impulse-momentum approach applied to the acceleration curve served to indirectly validate the accuracy of the film predicted forces imparted by the trampoline bed.

In addition to validating using impulse-momentum, Vaughan also compared the film measured displacement, velocity and acceleration curves to those theoretically predicted from a mathematical model. The trampoline bed was modelled to perform like an ideal spring, and theoretical equations were developed to predict displacement, velocity and acceleration for both the airborne and contact phases of the motion. The theoretically produced curves matched the experimental data extremely well, giving credibility to both of these sources. Vaughan (1980) reported that minor deviations in the results most likely occurred because the trampoline bed did "not behave in a perfectly elastic fashion", indicating that some damping was most likely occurring. No actual force history data from either the film (experimental) or the model (theoretical) was presented in this study. Although the acceleration curves from

both sources were compared, it would have been interesting to see a comparison of the force and joint moment curves produced from both of these methods.

An investigation by Patriarco et al. (1981) on human gait compared the joint moments of force obtained from film and force plate data to those calculated using film displacement data only. This comparison was part of an attempt to "evaluate the significance of various factors which contribute to the formation of a muscle force optimization solution" (Patriarco et al., 1981). Three-dimensional kinematic data was obtained by filming subjects with three perpendicularly arranged 16mm Photosonics cameras. An automated motion analysis system, TRACK (The Real time ACquisition of Kinematic data), which incorporates Selspot detectors with a computer based system, was used to verify the accuracy of the cinefilm data. Anthropometrics were taken from Braune and Fischer (1889), and a "simple Butterworth" (their term) filter was used for the purpose of data smoothing (Patriarco et al., 1981). The sampling frequency of the Photosonics cameras was not specified and the cinefilm displacement data, not the TRACK, was used for the joint moment calculations. In the calculation procedure for joint moments without force plate data, "the body was modelled as a five link system with force and moment equations for each limb" (Patriarco et al., 1981). Forces and moments in three dimensions were calculated for the ankle, knee and hip joints. Although no quantitative comparisons were made, visual inspection revealed good agreement between the general pattern of the film derived and force plate calculated vertical force curves, with only slight differences occurring in the magnitude of the results. Joint moments for flexion and extension (about the x -axis), were roughly similar in pattern (best for the

knee joint), but did not seem to be at a level of accuracy which would produce much confidence in the validity of the film derived data.

Problem areas which may have introduced error were, the validity of the anthropometric data chosen (both subjects used in this study were female), the appropriateness of a five segment model for deriving the joint kinetics, and the cutoff and sampling frequencies used, which were not specified with the other data collection and processing procedures. This study was useful in that it directly compared two methods (one with and one without force plate data) for calculating the joint forces and moments occurring in walking, for the same subject and the same trial. Relating to the primary purpose of the investigation, Patriarco et al. (1981) concluded that "the precision achieved in calculating joint torques dominates the muscle force distribution and is more influential in predicting muscle force activity than the mathematical techniques and assumptions used to compensate for muscle redundancy" (i.e. the particular optimization criteria employed). This statement was based on the finding that "a dramatic improvement" in the predicted temporal muscle pattern was achieved by replacing the joint moments calculated from film, with those obtained using the force plate data (Patriarco et al., 1981).

In the same year, Zarrugh (1981) computed ground reaction forces and joint moments for the single and double support phases of walking on a level treadmill, using absolute kinematic data calculated from relative motion measurements. A seven link rigid body model was used, which included a segment for the head-arms-trunk, along with thighs, shanks and feet for both of the lower limbs. The displacement data was collected using string transducers and self-aligning three-dimensional electrogoniometers sampling at 200 Hz,

and was then filtered by harmonic truncation at the twelfth harmonic (this technique reported in Zarrugh and Radcliffe, 1979). With respect to the predicted joint moments, Zarrugh (1981) states that "the ankle moments about the lateral axis (plantar/dorsiflexion) show excellent agreement with the results of Bresler and Frankel (1950)". The results of Bresler and Frankel (1950) are not presented in this paper, and no mention of similar agreement between knee and hip moments is referred to by Zarrugh. A visual comparison of the joint moment histories from both papers reveals a similar pattern between the lateral axis moments of the ankle, but in addition reveals poor comparisons for both knee and hip moments occurring about the same axis. Contrary to what was stated, joint moment results in this study do not show excellent or even satisfactory agreement with the findings of Bresler and Frankel (1950). One reason for the discrepancy of the results could be that walking on solid ground and walking on a treadmill are not perfectly identical or compatible skills. Considering that the ground is not behaving in the same fashion, it is quite conceivable that the joint moments for these two skills should in fact be somewhat different.

In a study which examined the GRF characteristics present in a gymnastics skill (the running forward somersault), Miller and Nissinen (1987) also investigated whether or not acceptable GRF time histories for sports skills involving rapid changes in motion, could be reconstructed from film records in the absence of a force platform. Subjects (eight male gymnasts) were filmed using a 16mm Locam camera, perpendicularly positioned at the side of a Kistler force platform upon which the support phase of the somersault was executed. Displacement and force data were sampled at between 196 and 197 and 1000 Hz respectively. Body segment parameters were taken from Dempster (1955) and Dempster and

Gaughran (1967), and a seven link rigid body model was used to analyze the gymnasts' motion. Smoothing of displacement curves was achieved with a symmetric 2nd order Butterworth digital filter, using frequency cutoffs ranging between 7 and 9 Hz.

Differentiated twice from the displacement data, the acceleration of the total body centre of gravity (CG) was used to compute the vertical and horizontal GRF components from film.

These results were then compared to the histories obtained from the force platform.

Miller and Nissinen (1987) reported that the film generated GRF curves "were unable to duplicate the rapid rise and decay of the impact peak and were not particularly successful in reproducing the horizontal GRF time histories". A potential source of error may have been the discrepancy in sampling rates between the camera and force platform, which resulted in the impact phase of the skill being recorded by a very small number of frames of film (two or three). Miller and Nissinen (1987) also state that as a direct result of the smoothing process (especially at relatively low cutoff frequencies around 7 Hz), "faithful portrayal of the rapidly changing force magnitudes" in this type of skill was not permitted. In addition, it was postulated that error might be attributed "to the inability to accurately locate the coordinates of the proximal and distal ends of the trunk and possibly also to some degree of thoracic flexion" (Miller and Nissinen, 1987). Errors in locating the CG of the trunk would "significantly influence the determination of the total body CG position, which serves as the basis for the series of calculations required to predict the ground reaction force" (Miller and Nissinen, 1987). It is interesting to note that even though ground reaction forces were predicted with and without force platform data, lower extremity joint moments were not compared or even computed for this investigation. Miller and Nissinen (1987) concluded

that "although film is useful in providing position data, it is unable (at least using the methods employed in the present study) to provide the basis for generating sufficiently accurate GRF data for use in determining resultant joint moment time histories in skills that involve large impact forces and rapid changes in motion". With the vertical forces at impact averaging 13.6 times greater than body weight for this particular skill, it is quite conceivable that better results (in predicting the impact portion of the force curve) could be achieved for a movement containing a less extreme vertical impact, along with higher frequencies for camera speed and filtering cutoffs and a modified trunk segment for the rigid body model. The force histories presented in this paper do indicate (visually at least), that the non-impact portions of the GRF curve predicted from film data compare well with those obtained using a force platform.

An interesting, but somewhat different, approach to the validation of kinetic information obtained from kinematic data was that of Ladin and Wu (1991), who instrumented a compound pendulum so that force information could be determined using three separate methods. These methods included a system which combined accelerometer and position data, a system which used double differentiated position data only, and a strain gauge system which measured the three-dimensional forces caused by the motion of the pendulum. The forces derived from the two kinematically based systems were compared, using the information from the strain gauge system as the criterion. The kinematic data was obtained using a WATSMART optoelectronic system with two infrared-sensitive cameras operating at 100 Hz, and a triaxial accelerometer operating at 300 Hz. Acceleration information obtained from the WATSMART position data was filtered with a 3rd order, dual

pass Butterworth digital filter at cutoff frequencies of 5 and 15 Hz. The strain gauge and accelerometer data was left unfiltered. It was apparent that the displacement-time history of the WATSMART system for the z-direction, was noisier than either the x or the y-directions. It is possible that the cameras might have had some difficulty in recording motion in the z-direction, because of the fact that the LED markers were flatly fixed to the pendulum in a single plane. Force history results revealed excellent agreement for both the accelerometer and film derived forces, when compared to the strain gauge criterion. The film derived force curve filtered at 5 Hz was superior in quality to that filtered at the 15 Hz cutoff, due to the elimination of irrelevant noise. This is not surprising considering the primary frequency component of the pendulum force spectrum, as measured by the strain gauge system, was lower than 2 Hz in all of the three directions.

Although Ladin and Wu (1991) advocate their method of combining accelerometer and position data as an improvement which would "increase the frequency range of useful joint lead estimates", it is likely that in actual human motion studies this method would be plagued by the problem of skin motion artifacts (also a problem in collecting data from film), and would be less feasible than is suggested in this particular analysis. More importantly, for the simple swinging motion of a compound pendulum, Ladin and Wu (1991) have shown that kinetic information with a high degree of accuracy can be predicted from the differentiation of position data alone. Although these results are not necessarily specific to human locomotion, they do confirm the ability of two kinematic methods to accurately predict the kinetics associated with non-contact motion.

In a study similar to that of Miller and Nissinen (1987), Sanders et al. (1991) investigated "the accuracy with which ground reaction forces in activities involving impact could be derived". Four subjects (two male and two female) performed 10 drop jumps from a height of 40 cm above the ground. The drop jump involved landing on a force platform and then, immediately following a crouch, jumping as high as possible. Force data were sampled at 500 Hz using a Kistler force platform, and displacement data was collected at 200 Hz with a high speed video camera. The body segment parameters of the subjects were computed by the elliptical zone method of Jensen (1976), as opposed to the use of typical cadaver data. A nine segment rigid body model was used which was comprised of the head and neck, upper trunk, lower trunk, along with right and left arms, forearms, hands, thighs, shanks and feet. Left and right limbs were modelled as single segments "in view of the similarity of mass distribution of right and left sides and the symmetry exhibited in the jumping performances" (Sanders et al., 1991). To determine the vertical component of the GRF, the vertical acceleration component of the total body CG was derived from position coordinates using central difference formulas.

Two approaches were used in analyzing the various GRF time histories produced for this activity. The first compared the video derived GRF curves, filtered at frequency cutoffs ranging from 4 to 9 Hz, to the unfiltered directly measured GRF curve. The other approach compared both curves filtered at a cutoff frequency of 5 Hz. The directly measured force function was filtered at this frequency "in order to separate the low frequency active component from the high frequency component associated with impact" (Sanders et al., 1991). It was obvious that the greatest source of error in deriving the GRF time histories

from displacement data, was due to the inability of this method to accurately reproduce the high frequencies that are associated with impact. It was shown, however, that the video generated force curves provided an almost perfect match to the directly measured force data when both were filtered at cutoff frequencies of 5 Hz. Based on these results it was concluded that, in a drop landing jump from a height of 40 cm, the low frequency part of the vertical force curve (< 5 Hz) could be estimated to within 3% of the maximum force, using the inverse dynamics approach calculated from positional data only.

Not long after the Sanders et al. (1991) paper was published, Bobbert et al. (1991) examined the segmental contributions to the vertical GRF impact peak determined from positional data in the landing phase of running. To evaluate the accuracy of the positional data method, time histories of the sum of the segmental contributions were compared to the vertical GRF measured directly using a Kistler force platform. The purpose was to establish whether or not the calculation of segmental contributions to the vertical GRF (F_z) component at impact was a more suitable approach than measuring F_z only, in seeking to examine the specific relationship between running injuries and impact force peaks.

A seven segment linked rigid body model was employed which included both feet, lower legs and thighs, along with a segment representing the head, arms and trunk (HAT). Body segment parameters were taken from Clauser et al. (1969) and applied to three male subjects, who performed a variety of different running speeds and styles in order to obtain a broad spectrum of GRF time histories. Force data were sampled at 1000 Hz, with the positional data collected from four electronically shuttered video cameras operating at 200 Hz. The video data was smoothed using a 2nd order zero lag Butterworth digital filter at

various cutoff frequencies, using padding points of 15 frames in length on either side of the data to eliminate filter boundary effects.

In an attempt to ameliorate the problem of skin movement relative to the underlying tissue, a device consisting of two light wooden rods connected by a hinge joint was constructed. The hinge joint was aligned with the flexion/extension axis of the knee, with one end fastened to the lower leg and the other to the upper leg. Markers were then placed on the rods at the respective joint axes of rotation, with the purpose of this system being to keep the distance between the markers constant. Although this modification in trying to improve the accuracy of recording positional data is conducive to a laboratory environment, it would not be possible to implement this method in recording the positional data of runners in a competition (or other possible field situations). In addition, it was not discussed whether or not any infringement was made by the wooden rods on the normal running style of the subjects.

Results of the comparison between the derived and the directly measured GRF curves indicated that when using a filter cutoff frequency of 50 Hz, extremely large fluctuations occurred in the derived curve as expected. With the cutoff reduced to 10 Hz the fluctuations of the derived curve disappeared, including the vertical impact force peak. A good result in reproducing the impact peak was achieved by using a combination of cutoff frequencies. These included 50 Hz for the foot, ankle and knee markers, 20 Hz for the hip markers and 15 Hz for the markers at the sternum and the head. The rationale for using an intermediate cutoff frequency for the hip markers, was that they are used in the acceleration calculations of both relatively light (lower legs) and heavy (HAT) body segments. Consequently, it was

concluded that an important factor with respect to accurately reproducing the impact portion of the vertical GRF curve, was the selection of a high frequency cutoff for the leg segment markers along with a low frequency cutoff for the other markers. Overall, the magnitude of the impact force peak was predicted from the positional data with errors of less than 10%. It was also established that the setup of four cameras for video data collection was unnecessary. The data obtained from a single lateral camera produced "reasonably accurate force calculations" (Bobbert et al., 1991).

It is evident from the preceding section that numerous researchers have investigated the possibility of predicting human movement force or moment information from displacement data. These studies have employed different activities (with either large or small impact forces), different displacement data collection methods, different body segment parameters and different linked segment models in examining this possibility. Different emphasis has also been placed on the particular aspect (forces or moments) of the movement kinetics analyzed. It seems acceptable that vertical ground reaction force time histories can be accurately predicted from displacement data, provided that the impact component of the skill is not excessively large. It remains to be seen whether or not sufficiently accurate net joint moments of force, for a medium impact oriented activity such as running, can be determined from displacement data using force platform calculated moments of force as the criterion for error quantification. It is hoped that if joint moments can be accurately predicted in this fashion it will be possible to analyze the kinetics of human motion in restricted situations where the use of a force platform is not possible.

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