Pergamon

J. Biomechanics, Vol. 28, No. 10, pp. 1243–1256, 1995
Copyright © 1995 Elsevier Science Ltd
Printed in Great Britain. All rights reserved
0021-9290/95 $9.50 + .00

0021-9290(94)00176-6

# TECHNICAL NOTE

## A COMPARISON OF INTERSEGMENTAL JOINT DYNAMICS TO ISOKINETIC DYNAMOMETER MEASUREMENTS

Kenton R. Kaufman,* Kai-Nan An† and Edmund Y. S. Chao‡

*Motion Analysis Laboratory, Children's Hospital, San Diego, CA 92123; †Orthopedic Biomechanics Laboratory, Department of Orthopedics, Mayo Clinic/Mayo Foundation, Rochester, MN 55905; and ‡Orthopedic Biomechanics Laboratory, Department of Orthopedics, Johns Hopkins University, Baltimore, MD 21205, U.S.A.

**Abstract**—This study quantified the errors associated with measurements obtained using an isokinetic dynamometer for knee evaluation at 60° and 180° $s^{-1}$. Five normal subjects were tested. A triaxial electrogoniometer was used to measure the kinematics. A custom-designed load cell was used to measure the three orthogonal components of the reaction force applied by the isokinetic dynamometer to the leg. The inverse dynamics problem was solved to obtain the intersegmental dynamics at the knee joint. These data were compared to those obtained from the isokinetic dynamometer. The results indicate that the knee motion recorded by the isokinetic dynamometer was in error by 7–9° during flexion and 10–13° during extension. The isokinetic dynamometer recorded knee flexions which were less than actual during the flexion portion of the exercise and greater than actual during extension. The duration of constant velocity was limited based on the speed selected. The knee moment was in error by 5 Nm during flexion and 25–29 Nm during extension. The use of a gravity correction factor reduced the knee moment error during extension, but increased it during flexion.

*Keywords*: Isokinetic dynamometer; Knee; Kinematics; Kinetics; Inverse dynamics.

### NOMENCLATURE

| | |
|---|---|
| $\mathbf{A}_j$ | externally applied force vector: $A_1$ is superior–inferior shear, $A_2$ is anterior–posterior shear, $A_3$ is medial–lateral shear |
| $B_i$ | body segments representing the lower extremity |
| c | cosine |
| **cg** | location of the center of mass from the point of body segment rotation |
| $[D]$ | matrix of direction cosines of the principal axes with respect to the anatomical axes |
| $\mathbf{F}_C$ | resultant force at body center of mass |
| $\mathbf{F}_D$ | force reaction constraint of the distal joint |
| $\mathbf{F}_P$ | force reaction constraint of the proximal joint |
| **g** | locally defined gravitational constant |
| $\mathbf{H}_A$ | angular momentum of the body at point A |
| $\dot{\mathbf{H}}_A$ | rate of change of angular momentum at point A |
| $[I]$ | inertia tensor about any point |
| $[I_C]$ | centroidal inertia tensor in the anatomical reference frame |
| $[I_P]$ | centroidal principal inertia tensor |
| **i, j, k** | components of vector in the distal coordinate system |
| **I, J, K** | components of vector in the proximal coordinate system |
| $\mathbf{l}_D$ | location of $\mathbf{F}_D$ |
| $m$ | body-segment mass |
| $\mathbf{M}_A$ | resultant moment vector acting on the free body at point A |
| $\mathbf{M}_D$ | moment reaction constraint of the distal joint |
| $\mathbf{M}_i$ | externally applied couple |
| $\mathbf{M}_P$ | moment reaction constraint of the proximal joint |
| $\mathbf{r}_i, \dot{\mathbf{r}}_i, \ddot{\mathbf{r}}_i$ | position, velocity, and acceleration of point $i$ in the inertial coordinate system |
| s | sine |
| $\mathbf{s}_j$ | position vector of $\mathbf{A}_j$ |
| $x, y, z$ | location of the mass center with respect to a parallel coordinate system |
| $\theta, \dot{\theta}, \ddot{\theta}$ | ab/adduction angle, velocity, and acceleration; adduction is positive |
| $\boldsymbol{\rho}_i, \dot{\boldsymbol{\rho}}_i, \ddot{\boldsymbol{\rho}}_i$ | position, velocity, and acceleration of point $i$ in the local coordinate system |
| $\phi, \dot{\phi}, \ddot{\phi}$ | flexion–extension angle, velocity, and acceleration; flexion is positive, |
| $\psi, \dot{\psi}, \ddot{\psi}$ | internal–external rotation, velocity, and acceleration; internal rotation is positive |
| $\omega, \dot{\omega}$ | angular velocity and acceleration of the moving body with respect to the inertial frame |

*Received in final form* 17 *November* 1994.

Address correspondence to: Kenton Kaufman, Ph.D., Motion Analysis Laboratory – 5054, Children's Hospital, 3020 Children's Way, San Diego, CA 92123, U.S.A.

### INTRODUCTION

Strength measurements are considered a core aspect of physical assessment. Isokinetic devices are commonly used to obtain strength measurements in functional testing and therapy programs. These devices are constant-velocity dynamometers which control movement by giving resistance at a preset speed of angular rotation. The measurements are restricted to a single plane. The widespread use of these measurements appears to have imparted to them a credibility that far exceeds what can currently be supported by research. It has been shown that there is a limited range of motion over which the movement is truly isokinetic while the

remainder of the range of motion is used for limb acceleration or deceleration (Osternig *et al.*, 1983b). Typically, the intersegmental joint moments are taken to be equal to the moments measured by the isokinetic dynamometer (Osternig *et al.*, 1983a; Perrine and Edgerton, 1978; Thorstenson *et al.*, 1976). Research has demonstrated that data from isokinetic dynamometers are error ridden and of questionable value (Grangsberg and Knuttson, 1983; Herzog, 1988; Sapega *et al.*, 1982; Winter *et al.*, 1981). Winter *et al.* (1981) demonstrated that an error in isokinetic dynamometry occurs if the gravitational forces are not taken into account. Sapega *et al.* (1982) performed movements in the horizontal plane in order to eliminate effects due to gravity. Gransberg and Knuttson (1983) corrected for the large oscillations in the torque reading during the early phase of fast isokinetic movements by using resistance to control the acceleration. Herzog (1988) compared differences in the movements measured by an isokinetic dynamometer to the calculated resultant joint moment. Differences were shown due to gravitational effects, inertial effects, and nonrigidity of the Cybex arm–leg–foot system. Yet, the analysis by Herzog (1988) was simply a planar analysis and assumed the reaction force applied by the isokinetic dynamometer was perpendicular to the long-axis of the leg and in a single plane of motion. These assumptions may not be correct. Differences between the moments measured with an isokinetic dynamometer and the three-dimensional intersegmental moments may exist.

The purpose of this research was to (1) describe the three-dimensional kinematics and kinetics of the knee during isokinetic exercise and (2) compare these values to those recorded by the isokinetic dynamometer.

## METHODS

The lower extremity was modeled as a mechanical system composed of four rigid body segments, $B_i$, corresponding to the pelvis ($B_0$), thigh ($B_1$), leg ($B_2$), and foot ($B_3$) (Fig. 1). An axis system was fixed in each body segment and moved with it. These coordinate systems were used to define joint orientation. The coordinates of bony landmarks were used to construct a right-handed orthogonal anatomically based reference frame for each coordinate system (McConville *et al.*, 1980). The positive $x$-axis was oriented in the superior direction. The positive $y$-axis was oriented in the anterior direction. The positive $z$-axis extended from the origin towards the left of the body. The rigid body segments were assumed to be interconnected by a ball-and-socket joint (joint with three rotational degrees of freedom and no translation). Although joint translation does occur in normal and pathological joints, its magnitude is usually small and difficult to measure (Chao, 1980). Consequently, the lines of action of the joint reaction forces were considered as passing through the joint centers, and the joint moments calculated were the net reaction moments needed to sustain motion and generate the measured moment. The coefficient of friction between articulating surfaces is very low, i.e. 0.001–0.005 (Andersson and Schultz, 1979; Radin and Paul, 1972; White and Panjabi, 1978). Accordingly, the articulating surfaces between bodies were assumed to be free of friction.

A unified definition of joint motion based on the three-axis Eulerian angles system (Chao, 1980; Grood and Suntay, 1983) was used in this study. The flexion–extension angle ($\phi$) occurred about a lateromedially directed axis fixed to the femoral condyle, and axial rotation ($\psi$) was measured about an axis along the shaft of the tibia (Fig. 2). The third axis (also defined as the floating axis) was orthogonal to the first two axes and defined abduction–adduction ($\theta$). Thus, the coordinates of a point or the components of a vector, as defined in either axis system, could be related by the transformation equation:

$$\begin{bmatrix} i \\ j \\ k \end{bmatrix} = \begin{bmatrix} c\theta c\phi & c\theta s\phi & -s\theta \\ -c\psi s\phi + s\psi s\theta c\phi & c\psi c\phi + s\psi s\theta s\phi & s\psi c\theta \\ s\phi s\psi + c\psi s\theta c\phi & -s\psi c\phi + c\psi s\theta s\phi & c\psi c\theta \end{bmatrix} \begin{bmatrix} I \\ J \\ K \end{bmatrix}. \quad (1)$$

However, it is important to recognize that two of the rotational axes in this system are nonorthogonal when the joint departs from its neutral position. Consequently, the system is difficult to use for kinetic analysis. Angular velocity and acceleration have to be transformed into a set of inertial axes in terms of the Eulerian angles defined.

The angular velocity of the moving system with respect to the fixed system can be expressed as the vectorial sum of three partial angular velocities corresponding respectively to flexion–extension, abduction–adduction, and internal–external rotation. The angular velocity is

$$\boldsymbol{\omega} = \dot{\phi}\mathbf{K} + \dot{\theta}\mathbf{j}' + \dot{\psi}\mathbf{i}'', \quad (2)$$

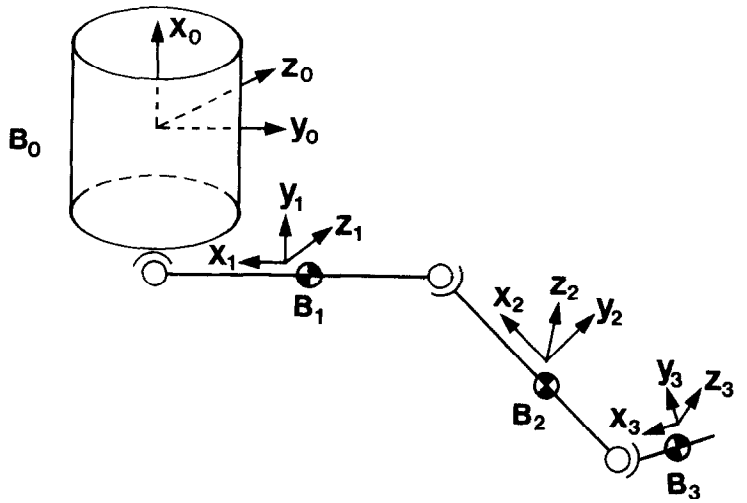



Fig. 1. Mechanical model of human lower limbs. Cartesian coordinate systems are defined in each body segment. The $x$-axis is positive in the proximal direction. The $y$-axis is positive in the anterior direction. The $z$-axis is positive to the left.

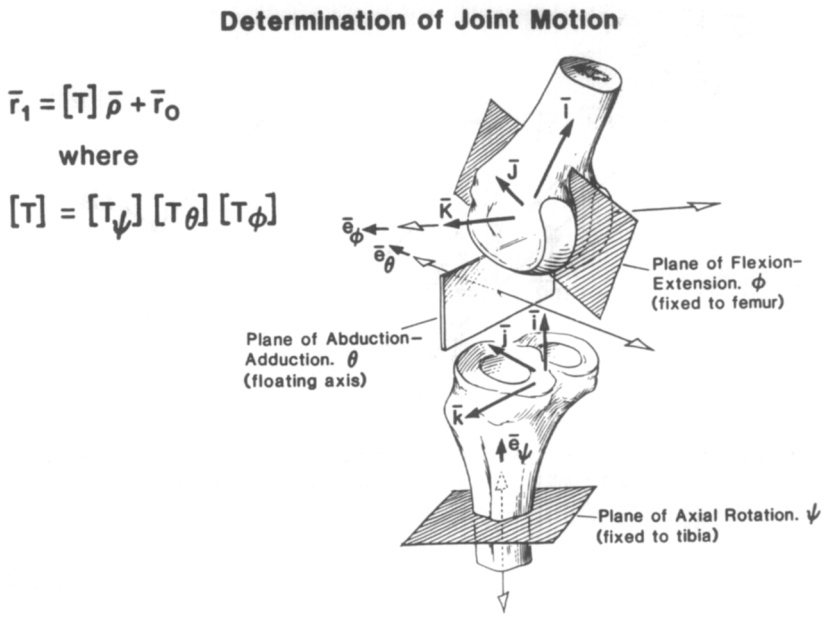



Fig. 2. Description of motion based on Eulerian angles. The proximal coordinate system is fixed to the distal femur and is denoted by capital letters, **I, J, K**. The distal coordinate system is fixed to the proximal tibia and is denoted by lower case letters, **i, j, k**. The *K*-axis ($\mathbf{e}_\phi$) is used to define the flexion–extension motion. The *i*-axis ($\mathbf{e}_\psi$) describes the internal–external rotation. The floating axis ($\mathbf{e}_\theta$) is orthogonal to both axes $\mathbf{e}_\phi$ and $\mathbf{e}_\psi$ and is used to measure the abduction–adduction motion. These three axes are mutually orthogonal when the knee joint is in its neutral position (zero rotation).

where the prime and double prime represent intermediary axis systems. Since the vector components obtained for $\boldsymbol{\omega}$ in equation (2) are not orthogonal, the unit vectors **K**, **j**′, and **i**″ must be resolved into components along the unit vectors of the rotating axes. The angular velocity with respect to the rotating axes in terms of the Eulerian angles is

$$\boldsymbol{\omega} = [\dot{\psi} - \dot{\phi}\sin\theta]\mathbf{i} + [\dot{\theta}\cos\psi + \dot{\phi}\cos\theta\sin\psi]\mathbf{j} + [\dot{\phi}\cos\theta\cos\psi - \dot{\theta}\sin\psi]\mathbf{k}. \quad (3)$$

The angular acceleration components; $\dot{\omega}_x$, $\dot{\omega}_y$ and $\dot{\omega}_z$; can be calculated directly from the angular velocity components, $\omega_x$, $\omega_y$, and $\omega_z$. After taking their time derivatives, the angular acceleration components become:

$$\begin{aligned}
\dot{\omega}_x &= \ddot{\psi} - \ddot{\phi}\sin\theta + \dot{\phi}\dot{\theta}\cos\theta,\\
\dot{\omega}_y &= \ddot{\theta}\cos\psi - \dot{\theta}\dot{\psi}\sin\psi + \ddot{\phi}\cos\theta\sin\psi\\
&\quad - \dot{\phi}\dot{\theta}\sin\theta\sin\psi + \dot{\phi}\dot{\psi}\cos\theta\cos\psi,\\
\dot{\omega}_z &= \ddot{\phi}\cos\theta\cos\psi - \dot{\phi}\dot{\theta}\sin\theta\cos\psi\\
&\quad - \dot{\phi}\dot{\psi}\cos\theta\sin\psi - \ddot{\theta}\sin\psi - \dot{\theta}\dot{\psi}\cos\psi. \quad (4)
\end{aligned}$$

The volume and principal moments of inertia of each body segment were calculated from various anthropometric dimensions using regression equations (McConville *et al.*, 1980). The center of volume and the principal axes were located with respect to the anatomical axes for each body segment (McConville *et al.*, 1980). The mass of the calf and foot were calculated by assuming the density of each body segment was 1.085 g cm$^{-3}$ (Clauser *et al.*, 1969). The location of the center of volume for each subject was scaled based on the length of their body segments in proportion to the average length of the body segments (McConville *et al.*, 1980) using the constant strain scaling method (Lewis *et al.*, 1980).

In order to obtain the inertia matrix corresponding to joint centers, it was necessary to rotate and translate the centroidal principal moments of inertia. The moments of inertia and products of inertia of a body segment with respect to orthogonal axes aligned with the anatomical axes and having a common origin with the principal centroidal moments of inertia were determined as:

$$[I_C] = [D][I_P][D]^T. \quad (5)$$

Next, the moment of inertia about an axis parallel to the centroidal axis was calculated using the parallel-axis theorem:

$$[I] = [I_C] + m\begin{bmatrix} y^2+z^2 & -xy & -xz \\ -xy & x^2+z^2 & -yz \\ -xz & -yz & x^2+y^2 \end{bmatrix}. \quad (6)$$

The forces and moments at the knee were determined using Newtonian mechanics. The fundamental equations are:

$$\sum \mathbf{F}_C = m\ddot{\mathbf{r}}_C, \quad (7)$$

$$\sum \mathbf{M}_A = (\dot{\mathbf{H}}_A)_{O_{XYZ}} + m(\dot{\mathbf{r}}_A \times \dot{\mathbf{r}}_C). \quad (8)$$

In order to solve for the proximal force and moment at any limb segment, a decision was made to solve for the moments of the external forces about the joint center, A, instead of at the limb segment's center of mass as is usually done (Fig. 3). For this case, the moment equation will not contain the unknown reaction forces at the joint center. Because of this method of solving the equations of motion, the components of the angular momentum became

$$(\dot{\mathbf{H}}_A)_{O_{XYZ}} = m\boldsymbol{\rho}_{C/A} \times \ddot{\mathbf{r}}_A + m\boldsymbol{\rho}_{C/A} \times (\boldsymbol{\omega} \times \dot{\mathbf{r}}_A) + (\dot{\mathbf{H}}_A)_{O_{xyz}} + \boldsymbol{\omega} \times \mathbf{H}_A. \quad (9)$$

Under these circumstances, the right-hand side of equation (9) contained two additional terms to account for the fact that the joint was moving in space.

A free-body diagram of the lower leg was constructed. Once the time functions of the positional variables, inertial parameters, and the external reaction vectors were obtained for each model segment, the proximal intersegmental force

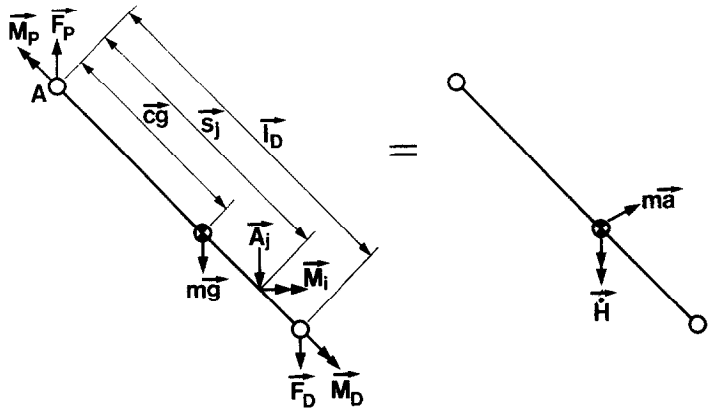



Fig. 3. Free body diagram of the leg under dynamic conditions. The system of external forces on the left are equipollent to the system of inertial forces on the right.

and moment vectors were calculated using the following equations:

$$\mathbf{F}_P = m(\ddot{\mathbf{r}}_O + \dot{\boldsymbol{\omega}} \times \mathbf{r}_C + \boldsymbol{\omega} \times (\boldsymbol{\omega} \times \mathbf{r}_C)) - \mathbf{F}_D - \sum_{j=1}^{n} \mathbf{A}_j - m\mathbf{g}, \quad (10)$$

$$\mathbf{M}_P = m\boldsymbol{\rho}_{C/A} \times \ddot{\mathbf{r}}_A + m\boldsymbol{\rho}_{C/A} \times (\boldsymbol{\omega} \times \dot{\mathbf{r}}_A) + (\dot{\mathbf{H}}_A)_{O_{xyz}} + \boldsymbol{\omega} \times \mathbf{H}_A + m(\dot{\mathbf{r}}_A \times \dot{\mathbf{r}}_C) - \mathbf{M}_D - \sum_{i=1}^{k} \mathbf{M}_i - \sum_{j=1}^{n} (\mathbf{s}_j \times \mathbf{A}_j) - \mathbf{l}_D \times \mathbf{F}_D - \mathbf{cg} \times m\mathbf{g}. \quad (11)$$

These equations were written with the vectors expressed in the coordinates of the local coordinate system located at the assumed center of rotation of the joint at the proximal end of the leg. The force and moment vectors for the foot calculated at the ankle center of rotation were transformed into the leg coordinate system. All reported forces and moments use the leg coordinate system as the reference frame. The forces are described in terms of their direction. The moments are named in terms of the reaction moment at the joint, e.g. an extension moment occurred during knee extension.

Five male subjects participated in this study. The age of the participants was $27 \pm 2$ (mean $\pm$ S.D.) years. They weighed $81 \pm 13$ kg, and had a height of $178 \pm 5$ cm. The right knee was tested for all subjects. To be included in the study, the subjects had to have no previous knee surgery and no history of knee disorders. Examinations revealed stable ligaments with a full range of knee motion symmetric with the other side, no effusions, and no generalized ligamentous laxity. Radiographic examination was used to rule out any bony abnormalities.

Angular displacement at the knee was measured with a triaxial electrogoniometer (Fig. 4). The triaxial electrogoniometer has been used in the measurement of three-dimensional rotations of the lower extremity joints during gait (Chao and Hoffman, 1978). Experimental and theoretical justifications have been performed with satisfactory results (Chao, 1980; Chao and Hoffman, 1978). The measurement error was less than two degrees for all three angular components (Chao, 1980). The potentiometer cluster was aligned with the knee rotation axis by a physical therapist familiar with the test equipment. Self-aligning mechanisms and instrument mounting guides were utilized to minimize error due to improper placement. Normal cross-talk due to external placement of the device and the noncollinear alignment of the axes of rotation was corrected mathematically (Chao, 1980). The goniometer was strapped to the thigh and leg with elastic straps and, with a mass of only 85 g, did not inhibit free movement.

A custom-designed, strain-gaged load cell was developed for this project (NK Biotechnical Engineering, Minneapolis, MN). This device was attached to the arm of the Cybex machine and accurately measured the applied force on the leg about three orthogonal axes (Fig. 4). The load cell was calibrated using a loading device traceable to the National Institute of Standards and Technology. The calibration scheme was linear with the output signal proportional to the applied load. The resolution of the load cell was $\pm 0.4$ N with a nonlinearity of less than 0.3% F.S. and a hysterisis of less than 0.45% F.S. Although cross-talk among all the channels was less than 1.0% F.S., a $3 \times 3$ correction matrix supplied by the manufacturer was used to eliminate completely the effect of cross-talk between the output channels.

A Cybex II isokinetic dynamometer was used to limit the angular velocity of the leg and provide resistance to movement (Fig. 4). Each subject was seated on a testing bench with the backrest reclined at 15° from the vertical. Two straps criss-crossed the chest of the subject to restrain the trunk against the back of the testing bench. A strap was placed across the anterior aspect of the distal end of the thigh to restrict motion of the thigh.

Each subject was instructed to exert maximal voluntary effort in both flexion and extension. Two speeds of 60 and 180° $s^{-1}$ were used. The higher speed was tested first. To ensure maximal effort, the movement was repeated until almost identical results were obtained. The subject then performed five trials at each speed without stopping. This number of trials was selected because it has been shown that at least three maximal trials are needed to obtain stable measures (Johnson and Siegel, 1978; Sawhill *et al.*, 1982).

Signals were acquired at 1 kHz per channel by a DEC microPDP 11/73 and transferred to a DEC VAX 11/750 for analysis. The generalized, cross-validatory spline smoothing and differentiation routine (Woltring, 1986) was used to estimate the angular velocity and acceleration from the angular displacements. The smoothing parameters were set automatically by the smoothing routine using statistical considerations to determine the proper amount of smoothing required (Craven and Wahba, 1979). The parameters were set to obtain a reasonable balance between goodness of fit to the raw data and smoothness of the fitted data in terms of the signal function and its derivatives (Woltring, 1985). The data were analyzed using the lower extremity model to obtain the kinematics and kinetics of the isokinetic exercise. The five trials for each subject were averaged to obtain a mean trial. The within subject variability was less than 3° for motion and less than 12 Nm for the intersegmental moment.

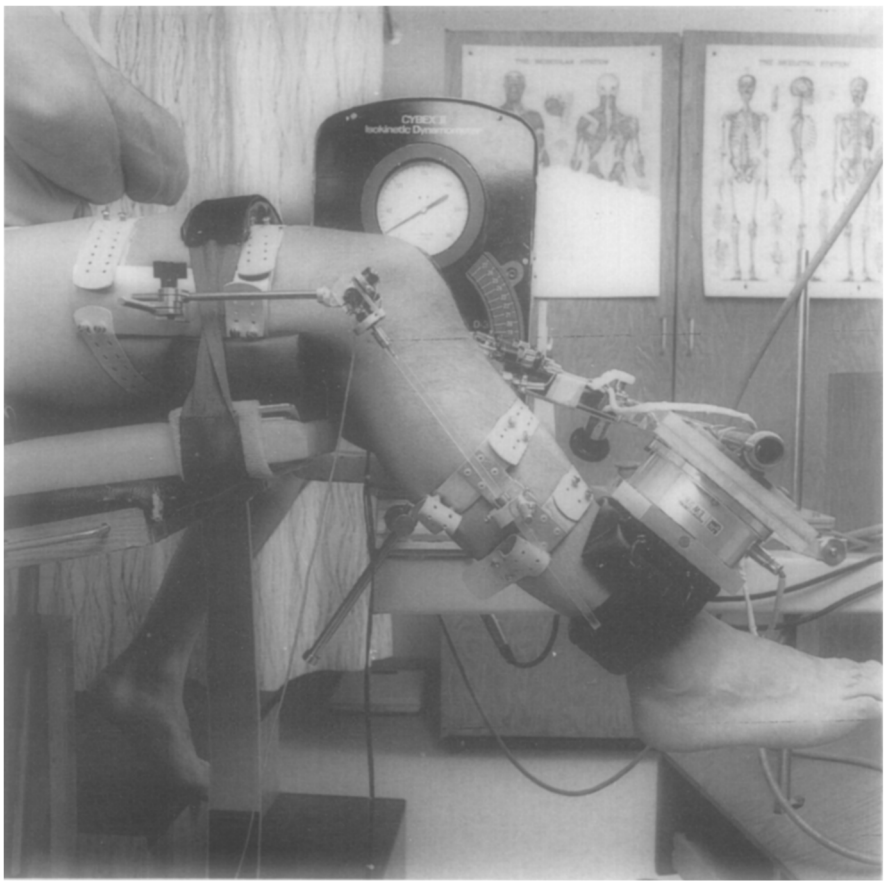


Fig. 4. Overview of test setup. The subject was seated on a bench which was inclined upward 15° from the horizontal. A triaxial electrogoniometer was mounted on the subject's lower extremity to measure the three-dimensional motion of the leg. A Cybex II isokinetic dynamometer was used as a loading device. A three component load cell was placed on the Cybex arm to measure the three orthogonal force components acting on the leg.

Statistical comparisons were made using a repeated measures ANOVA. The data were analyzed as a two factor design with repeated measures on both factors. The first factor was the measurements at each point of the exercise cycle. The second factor was the measurement technique. Separate analyses were performed for the flexion and extension portions of the exercise. The first and last 5% of each cycle portion was not included in the analysis to eliminate any spurious effects due to reversal of the movement.

## RESULTS

The motion of the leg recorded by the isokinetic dynamometer differed from the actual motion measured with a body-mounted goniometer (Fig. 5). During the flexion portion of the exercise, the machine measurement averaged $7° \pm 3°$ and $9° \pm 4°$ less than the true knee flexion at 60 and 180° $s^{-1}$, respectively. These differences were not significant ($p = 0.30$ at 60° $s^{-1}$ and $p = 0.11$ at 180° $s^{-1}$). During the extension portion of the exercise, the isokinetic dynamometer measurements were significantly greater than the true knee flexion by $13° \pm 2°$ at 60° $s^{-1}$ ($p = 0.01$) and $10° \pm 3°$ at 180° $s^{-1}$ ($p = 0.03$). The total range of motion was similar for both the body-mounted goniometer and machine goniometer. Using the body-mounted goniometer, the motion in all three planes was consistent for both speeds tested. The dominant motion was flexion–extension. However, there was motion in the other planes. The range of motion in all three planes was similar for both speeds tested (Table 1).

The angular velocities (Fig. 6) and angular accelerations (Fig. 7) showed that the regions of isokinetic movement were limited. At 60° $s^{-1}$, the region of constant velocity was during approximately 15–40% and 65–85% of the flexion–extension cycle. This is reflected in the fact that the angular acceleration is approximately zero in this range. In contrast at 180° $s^{-1}$, the angular velocity reached a value close to the established speed in approximately the first 5% of the flexion–extension cycle, but then the speed gradually drifted throughout the entire flexion motion. During extension at 180° $s^{-1}$, the angular velocity was constant only during the range 80–95% of the cycle. These unstable velocities were echoed by the fact that the angular accelerations were approximately zero during these time periods. The constant angular velocity differed from the preset speed because of the difference in angular position of the knee and the isokinetic dynamometer (Fig. 5).

The reaction force at the leg is not perpendicular to the lever arm of the isokinetic dynamometer (Fig. 8). There were transition regions at the extremes of the motion range where the force contributions varied. However, during the mid-range of flexion, i.e. 5–45% of the cycle, and extension, i.e. 55–95%, of the cycle the contributions were well defined at both speeds. The anterior–posterior contribution, which is perpendicular to the lever arm, accounted for 91–93% of the total force component during flexion. It was the dominant component at 95–98% during the early portion of extension, but decreased to 80–85% of the total force during terminal extension. There was a significant superior–inferior shear force (up to 15% of the total force component) during terminal extension. This additional force component contributes to the larger intersegmental moments which are not recorded by the isokinetic dynamometer. The medial–lateral shear was 4–6% of the total force component during flexion and 1–2% of the total force component during extension.

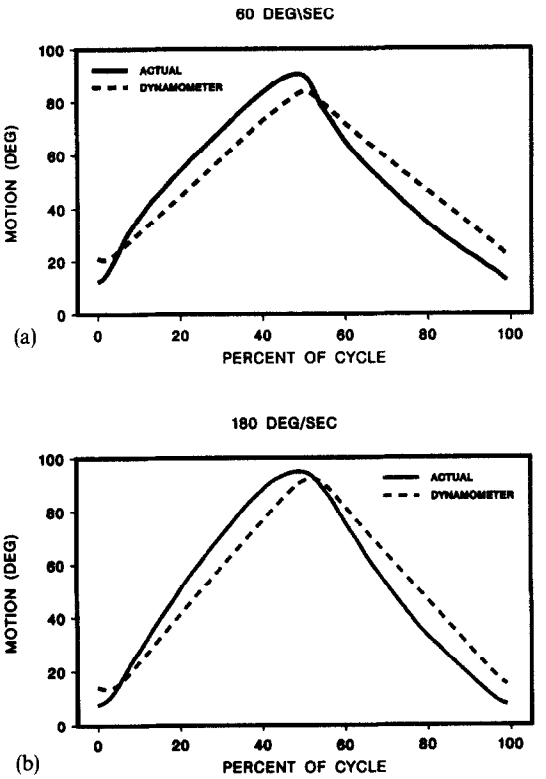
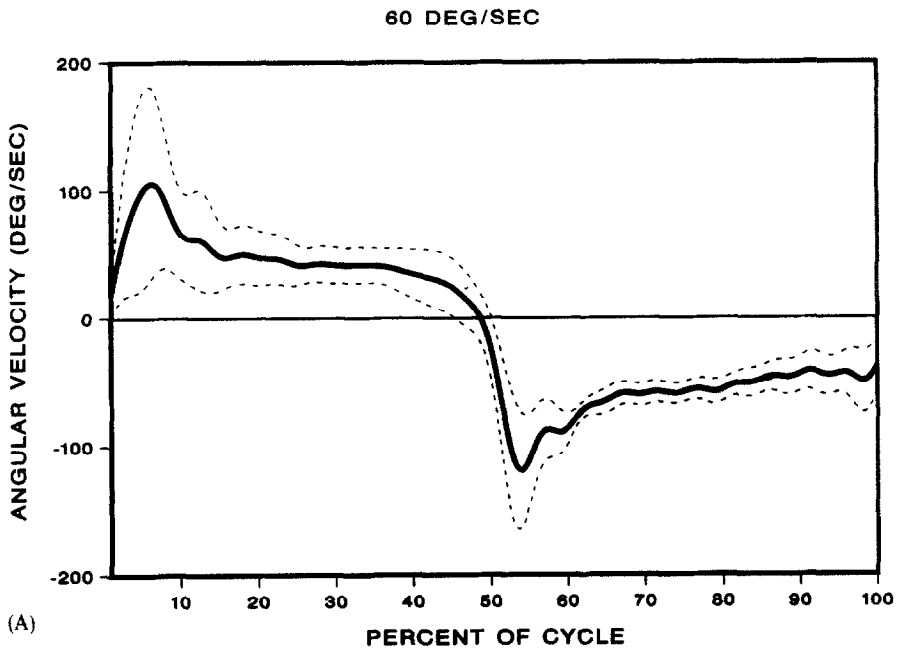



Fig. 5. Knee joint rotational motion during isokinetic exercise at (a) 60° $s^{-1}$ and (b) 180° $s^{-1}$. Full knee extension is recorded as zero degrees. The cycle starts with flexion and ends with extension. The solid line is the motion measured with a goniometer attached to the leg. The dashed line is the motion measured with an isokinetic dynamometer. The data are an average for five subjects. The standard deviation of the data is less than 7° for the goniometer and less than 10° for the isokinetic dynamometer.

Table 1. Knee motion during isokinetic exercise

| | 60° $s^{-1}$ | | | | 180° $s^{-1}$ | | | |
|---|---|---|---|---|---|---|---|---|
| | Minimum | | Maximum | | Minimum | | Maximum | |
| | Mean | S.D. | Mean | S.D. | Mean | S.D. | Mean | S.D. |
| Rotation | − 4.5 | 7.8 | 8.9 | 7.9 | − 4.0 | 9.9 | 6.3 | 7.8 |
| Ab/Ad | − 5.6 | 5.6 | 9.8 | 4.5 | − 3.4 | 2.9 | 10.0 | 7.1 |
| Flexion | 12.3 | 4.1 | 91.1 | 8.6 | 7.5 | 1.7 | 95.5 | 9.0 |

Internal rotation, adduction and flexion are positive.
$n = 5$.
S.D. = Standard deviation.



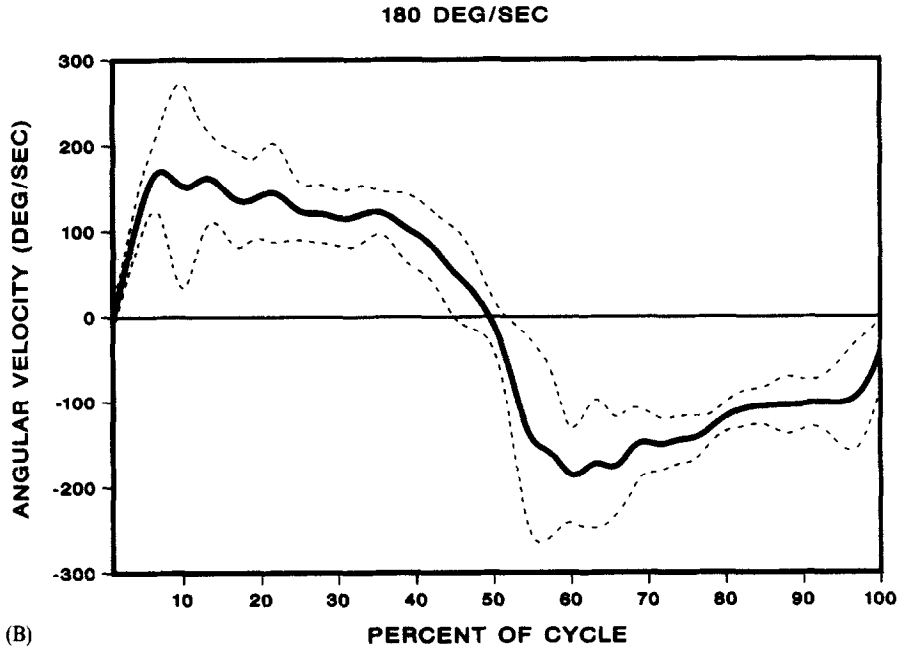




Fig. 6. Angular velocity during isokinetic exercise at (A) 60° $s^{-1}$ and (B) 180° $s^{-1}$. The cycle starts with flexion and ends with extension. Solid line is average for five subjects. Dashed line indicates ±1 standard deviation.

The moment recorded by the isokinetic dynamometer differed from the actual intersegmental moment (Fig. 9). For flexion at 60° $s^{-1}$, the actual moment was a nonsignificant 5 ± 6 Nm (7 ± 8%) greater than the isokinetic dynamometer moment ($p = 0.54$). At 180° $s^{-1}$, the isokinetic dynamometer moment was a nonsignificant 5 ± 5 Nm (17 ± 34%) greater than the actual moment ($p = 0.14$). In contrast during the extension portion of the isokinetic movement, the actual moment was significantly greater by 25 ± 5 Nm (20 ± 5%) at 60° $s^{-1}$ ($p = 0.02$) and by 29 ± 15 Nm (28 ± 14%) at 180° $s^{-1}$ ($p = 0.004$) than the dynamometer moment. The use of a gravity correction factor reduced the differences during the extension portion of the exercise, but increased them during the flexion portion of the exercise. After using the gravity correction during flexion, the error was 22 ± 3 Nm (28 ± 3%) at 60° $s^{-1}$ and 10 ± 11 Nm (20 ± 50%) at 180° $s^{-1}$. These differences were statistically significant ($p = 0.06$ at 60° $s^{-1}$ and $p = 0.04$ at 180° $s^{-1}$) as compared to when the gravity correction was not used. During extension, the error was 14 ± 9 Nm (10 ± 7%) at 60° $s^{-1}$ and 17 ± 21 Nm (13 ± 24%) at 180° $s^{-1}$. These differences remained statistically significant ($p = 0.05$ at

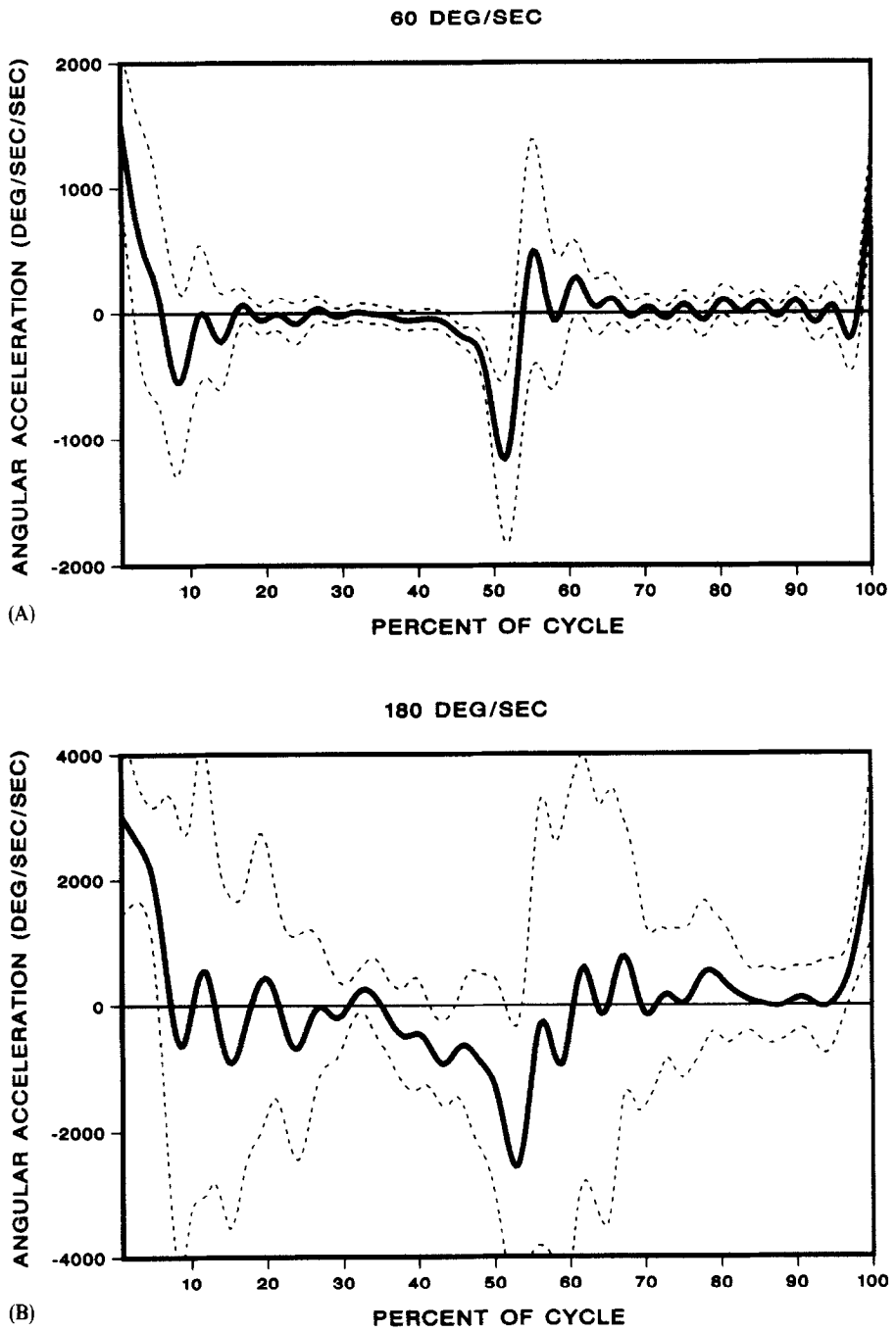
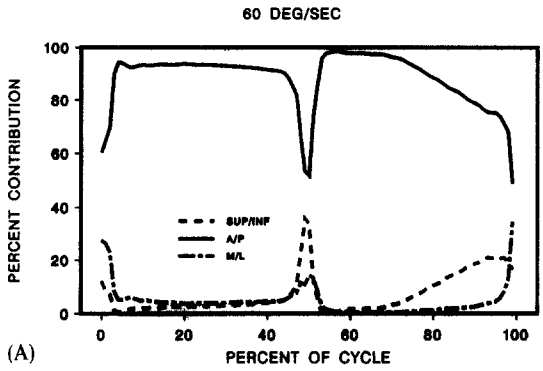



Fig. 7. Angular acceleration during isokinetic exercise at (A) $60^\circ s^{-1}$ and (B) $180^\circ s^{-1}$. The cycle starts with flexion and ends with extension. Solid line is average for five subjects. Dashed line indicates $\pm 1$ standard deviation.

$60^\circ s^{-1}$ and $p = 0.01$ at $180^\circ s^{-1}$) as compared to when the gravity correction was not used.

The maximum intersegmental forces and moments give an indication of the relative three-dimensional nature of the exercise. The dominant intersegmental reaction force occurred in the anterior–posterior direction as a result of the force being applied by the subject to the isokinetic dynamometer (Table 2). Posterior forces occurred during flexion. Anterior forces occurred during extension. Normalizing these forces to the subject's bodyweight, the axial force was found to be 0.19 and 0.15 BW at 60 and $180^\circ s^{-1}$, respectively. The anterior force was 0.63 and 0.52 BW while the posterior force was 0.34 and 0.22 BW for the speeds of 60 and $180^\circ s^{-1}$, respectively. The coefficient of variability was 0.4 for axial force, 0.2 for anterior–posterior forces, and 0.7 for medial–lateral forces. These values were the same regardless of speed of exercise or whether the forces were or were not normalized. The largest intersegmental reaction moment was the flexion–extension moment (Table 3). The abduction–adduction moment and the rotational moment at $180^\circ s^{-1}$ were also notable. The peak intersegmental moments mostly occurred at knee flexion angles from 40° to 65°.



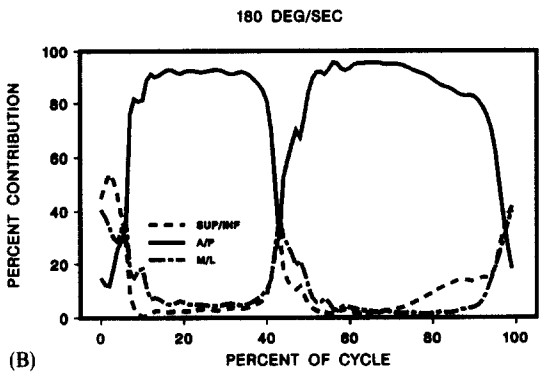
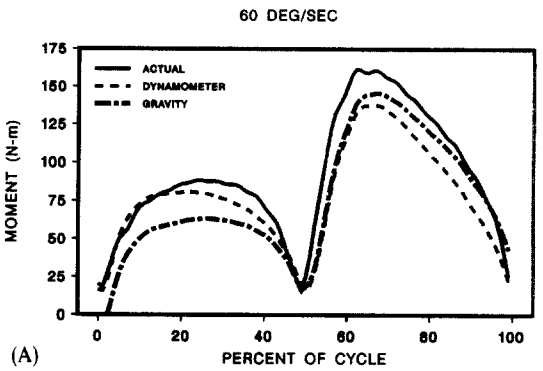




Fig. 8. Percentage contribution of orthogonal force components to the total reaction force supplied by the isokinetic dynamometer at (A) 60° $s^{-1}$ and (B) 180° $s^{-1}$. The cycle starts with flexion and ends with extension. The solid line is the contribution from the anterior–posterior force component. The dashed line is the contribution from the superior–inferior force component. The dash–dotted line is the contribution from the medial–lateral force component. The data are an average for five subjects. The standard deviation of the data is less than 5%, 10% and 5% for the anterior–posterior, superior–inferior, and medial–lateral force components, respectively.



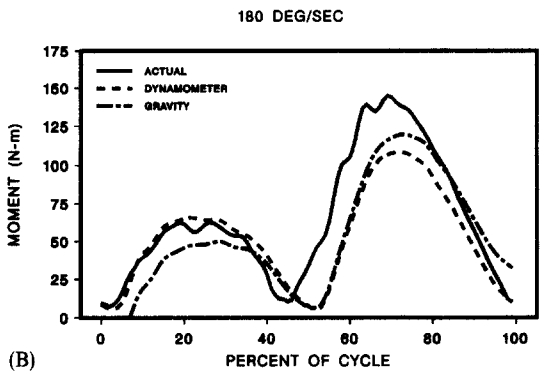




Fig. 9. Knee joint flexion–extension moments during isokinetic exercise at (A) 60° $s^{-1}$ and (B) 180° $s^{-1}$. The cycle starts with flexion and ends with extension. The solid line is the actual moment calculated with a model of the lower extremity. The dashed line is the moment recorded by the isokinetic dynamometer. The dash–dotted line is the isokinetic dynamometer moment after applying a gravity correction. The data are an average for five subjects. The standard deviation of the data is less than 30, 23 and 22 Nm for the actual, isokinetic dynamometer without gravity correction and with gravity correction, respectively.

Table 2. Maximum intersegmental forces during isokinetic exercise

| | 60° $s^{-1}$ | | | | 180° $s^{-1}$ | | | |
|---|---|---|---|---|---|---|---|---|
| | Force (N) | | Knee flexion (deg) | | Force (N) | | Knee flexion (deg) | |
| | Mean | S.D. | Mean | S.D. | Mean | S.D. | Mean | S.D. |
| Cephalid | 147 | 56 | 51 | 27 | 123 | 51 | 42 | 11 |
| Anterior | 494 | 111 | 56 | 9 | 413 | 66 | 54 | 7 |
| Posterior | 269 | 75 | 63 | 21 | 175 | 48 | 51 | 10 |
| Medial | 76 | 50 | 28 | 14 | 78 | 55 | 34 | 14 |
| Lateral | 18 | 39 | 57 | 23 | 37 | 34 | 62 | 8 |

$n = 5$.
S.D. = Standard deviation.

The relative contribution of the isokinetic dynamometer reaction force, gravity, inertia and the distal segment (foot) to the magnitude of the proximal moment was also examined (Fig. 10). The dominant component was the externally applied reaction force on the leg. The contributions of gravity, the distal segment and inertia were minor at both speeds. The inertial component was greater at 180° $s^{-1}$, but it was not a dominant component as might be expected. Also at the higher speed, the proportion due to inertia was more dominant during the initial part of both flexion and extension. During knee extension the externally applied force accounted for 90% of the proximal moment at 60° $s^{-1}$ and 85% at 180° $s^{-1}$ (Table 4). The other components contributed less than 5% with two exceptions. The contribution due to

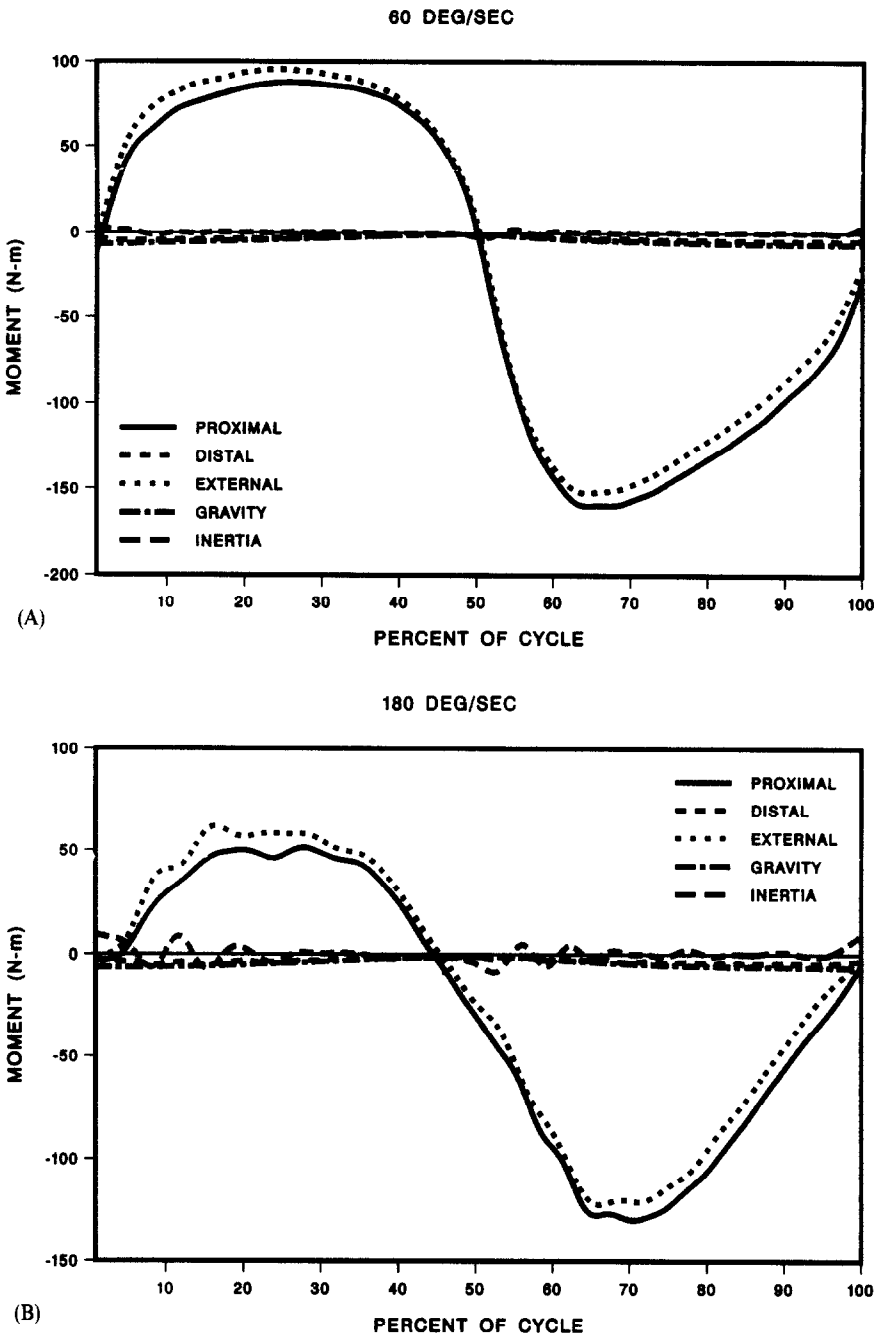



Fig. 10. Relative contribution of the distal force, gravity, externally applied reaction force and inertia to the proximal intersegmental flexion–extension moment at (A) 60° $s^{-1}$ and (B) 180° $s^{-1}$.

gravity at 60° $s^{-1}$ was 6% and the contribution from inertia at 180° $s^{-1}$ was 8%. During knee flexion, there was a difference in signs of some of the components. This is possible since the proximal moment is the algebraic sum of the respective components. The contributions due to the distal segment and gravity would aid flexion, whereas the reaction force and inertia would oppose flexion. Once again, the reaction force was the dominant component and contributed 113% of the proximal moment at 60° $s^{-1}$ and 112% at 180° $s^{-1}$. The contribution from the distal segment and gravity were in the opposite direction to the applied force and each contributed 6–9% of the net joint moment. The contribution from inertia was smaller than the other components and fairly constant at 3% of the net moment.

## DISCUSSION

Differences in knee flexion were observed for the body-mounted goniometer compared to the measurements reported by the isokinetic dynamometer. These differences may be explained by the fact that rarely, if ever, does the knee joint axis of rotation remain in alignment with the machine axis of rotation during dynamic isokinetic exercise. When

Table 3. Maximum intersegmental moments during isokinetic exercise

| | 60° s$^{-1}$ | | | | 180° s$^{-1}$ | | | |
|---|---|---|---|---|---|---|---|---|
| | Moment (Nm) | | Knee flex (deg) | | Moment (Nm) | | Knee flex (deg) | |
| | Mean | S.D. | Mean | S.D. | Mean | S.D. | Mean | S.D. |
| Int. Rot. | 5 | 5 | 42 | 23 | 14 | 13 | 38 | 12 |
| Ext. Rot. | 3 | 2 | 66 | 13 | 15 | 13 | 66 | 16 |
| Abduction | 15 | 7 | 66 | 15 | 23 | 16 | 66 | 11 |
| Adduction | 27 | 21 | 28 | 16 | 25 | 20 | 33 | 17 |
| Flexion | 93 | 28 | 63 | 21 | 70 | 30 | 48 | 7 |
| Extension | 166 | 41 | 56 | 9 | 154 | 38 | 54 | 7 |

$n = 5$.
S.D. = Standard deviation.

Table 4. Average percentage distribution of proximal intersegmental moment

| Motion | Speed (deg/s) | Distal | | Gravity | | Reaction force | | Inertia | |
|---|---|---|---|---|---|---|---|---|---|
| | | Mean | S.D. | Mean | S.D. | Mean | S.D. | Mean | S.D. |
| Extension | 60 | 4.5 | 1.0 | 5.9 | 1.2 | 89.8 | 1.3 | − 0.2 | 1.0 |
| Extension | 180 | 4.0 | 4.4 | 3.1 | 9.4 | 84.6 | 2.1 | 8.3 | 13.2 |
| Flexion | 60 | − 6.5 | 6.8 | − 9.3 | 11.6 | 113.1 | 10.0 | 2.7 | 11.2 |
| Flexion | 180 | − 6.1 | 14.6 | − 8.5 | 19.8 | 111.7 | 36.5 | 2.9 | 12.2 |

$n = 5$.
S.D. = Standard deviation.

the subject was setup for exercise, the knee joint axis of rotation was aligned with the machine axis of rotation. Even when care was taken to stabilize the limb with straps, the forces generated during the exercise resulted in movement of the joint relative to the machine axis. During flexion, the knee angle was greater than reported by the isokinetic dynamometer. Conversely, during extension the knee was extended to a greater degree than was measured by the machine. Herzog (1988) also observed that the knee flexion angle was smaller than the value recorded by the isokinetic dynamometer during extension exercise. The differences between the machine and the true knee position are in opposite directions which confirms that the knee axis of rotation is varying about the machine axis. These differences of 7°–13° may be important when joint angle specific measurements are being reported.

Another issue of importance in isokinetic measurement is the angular velocity of the movement. Although the term isokinetic literally means uniform motion, the duration of constant velocity is limited and related to the preset speed. The acceleration, oscillation and deceleration of the limb limits the range of constant velocity. Osternig *et al.* (1983b) reported a decreased isokinetic range with increased dynamometer speed settings. They observed that from 92% to 16% of knee extension at speeds from 50° to 400° s$^{-1}$, respectively, were in a constant velocity mode when the knee extension went from 120° to 0°. Lander *et al.* (1985) found that for an isokinetic bench press apparatus, the constant velocity portion was only 50% at 100° s$^{-1}$. Based on observations made in this study, it is apparent that periods of constant velocity do occur during isokinetic exercise. However, the duration of constant velocity is limited within boundaries determined by the speed of exercise and the capabilities of the individual being tested.

This study demonstrated differences in the moment readings of an isokinetic dynamometer and the actual intersegmental moment. All isokinetic dynamometers simply measure the moment at the axis of the machine. In the past, it has been assumed that this moment is equivalent to the resultant joint moment (Osternig *et al.*, 1983a; Perrine and Edgerton, 1978; Thorstenson *et al.*, 1976). In contrast, Herzog (1988) reported differences between the actual resultant joint moment and the moment recorded by the Cybex dynamometer. He studied knee extension only and reported errors ranging from 10 Nm (9.4%) at 120° s$^{-1}$ to 9.6 Nm (17%) at 240° s$^{-1}$. He also noted that the error was greatest at high flexion angles. His results are less than those reported in this study. The differences are partially explained by the shear loads at the leg which contribute to the intersegmental moment but which act parallel to the dynamometer lever arm and thus do not contribute to the machine moment.

The terminology of intersegmental loading is deliberately utilized to indicate the resultant loads transmitted between the thigh and leg, rather than the frequently used but confusing terminology 'knee force and moment' or 'net quadriceps–hamstring moment'. It is important to realize that the resultant intersegmental load and moment cannot be used to assess intrinsic function of the knee joint without dynamic muscle modeling. The term 'knee joint force' is generally reserved for the loading transmitted at the articulating surfaces of the knee. The resultant axial loading given in Table 2 should not be interpreted to be the net load transmitted between the articulating surfaces of the femur and tibia.

The maximum intersegmental moments during extension of the knee have been estimated for various activities of daily living (Table 5). The peak extension moments reported during isokinetic exercise are greater than the moments reported for cycling (Ericson and Nisell, 1986), level walking (Bresler and Frankel, 1950; Boccardi *et al.*, 1980; Cavanagh and Gregor, 1975; Morrison, 1968; Radcliffe, 1962; Winter, 1980), ascending stairs (Andriacchi *et al.*, 1980), rising from a squat (Schüldt *et al.*, 1983), or rising from a chair (Kelley *et al.*,

Table 5. Maximum moments during knee extension

| Author | Activity | Knee angle (deg) | Moment (Nm) |
|---|---|---|---|
| Present study | 60° $s^{-1}$ | 56 | 166 |
| | 180° $s^{-1}$ | 54 | 154 |
| Ericson and Nisell, 1986 | Cycling | 90 | 25 |
| Cavanagh and Gregor, 1975 | Walking (swing) | 20 | 25 |
| Winter, 1980 | Walking | — | 45 |
| Bresler and Frankel, 1950 | Walking | 20 | 47 |
| Morrison, 1968 | Walking | — | 51 |
| Andriacchi *et al.*, 1980 | Up Stairs | 65 | 54 |
| Boccardi *et al.*, 1981 | Walking | — | 60 |
| Radcliffe, 1962 | Walking | 20 | 61 |
| Schüldt *et al.*, 1983 | Rising from squat | 105 | 70 |
| Kelley *et al.*, 1976 | Rising from chair | 90 | 110 |
| Smidt, 1973 | Isometric | 45 | 119 |
| Knapik *et al.*, 1983 | Isokinetic, 180° $s^{-1}$ | 60 | 130 |
| Andriacchi *et al.*, 1980 | Down Stairs | 60 | 147 |
| Nisell, 1985 | Isokinetic, 180° $s^{-1}$ | 65 | 181 |
| Wahrenberg *et al.*, 1978 | Kicking a ball | 100 | 182 |
| Nisell, 1985 | Isometric | 90 | 198 |
| Winter, 1983 | Jogging | 50 | 210 |
| Lindahl *et al.*, 1969 | Isometric | 60 | 225 |
| Knapik *et al.*, 1983 | Isometric | 70 | 250 |
| Nisell, 1985 | Parallel squat | 120 | 455 |
| Zernicke *et al.*, 1977 | Tendon rupture | 90 | 550 |

Table 6. Maximum moments during knee flexion

| Author | Activity | Knee angle (deg) | Moment (Nm) |
|---|---|---|---|
| Present study | 60° $s^{-1}$ | 63 | 93 |
| | 180° $s^{-1}$ | 48 | 70 |
| Ericson and Nisell, 1986 | Cycling | 45 | 10 |
| Smidt, 1973 | Isometric | 60 | 52 |
| Knapik *et al.*, 1983 | Isokinetic, 180° $s^{-1}$ | 20 | 87 |
| Knapik *et al.*, 1983 | Isometric | 20 | 142 |

1978). They are comparable to descending stairs (Andriacchi *et al.*, 1980) and isokinetic extension exercise (Knapik, 1983; Nisell, 1985) reported in other studies. On the other hand, the maximum moments are less than for kicking a ball (Wahrenberg *et al.*, 1978a, b), jogging (Winter, 1983), or performing a parallel powerlifting squat (Nisell, 1985). The literature varies widely on the intersegmental moment for isometric extension, so no comparison can be made. Finally, the maximum intersegmental moments reported in this study are about 25–30% of those which caused rupture of the patellar tendon (Zernicke *et al.*, 1977).

The literature contains limited estimates of the intersegmental moments during flexion of the knee (Table 6). The peak flexion moments reported during isokinetic exercise are greater than the moments reported in the literature for isometric flexion (Smidt, 1973) or cycling (Ericson and Nisell, 1986). They are similar to those reported for isokinetic exercise in another study (Knapik *et al.*, 1983). The literature varies widely on the intersegmental moment for isometric flexion, so no comparison can be made.

When isokinetic dynamometers are used to record knee joint intersegmental moments (e.g. knee extensions and flexions), gravity either aids or opposes the movement. The percentage error tends to be large for less forceful contractions and decreases as the force of contraction increases (Winter *et al.*, 1981). Efforts are sometimes made to include a correction for gravity when using an isokinetic device. When the gravity correction was applied to the isokinetic dynamometer readings, the error decreased during extension. This agrees with the results reported by Herzog (1988). However, the gravity correction increased the error during flexion. The components of the actual intersegmental moment were also examined. The moment measured by the isokinetic machine is equivalent to the 'reaction force' component of the intersegmental moment (Table 4). The contribution of the distal segment, gravity and inertia were small in comparison to the reaction force. The fact that the externally applied force is the dominant component is important clinically since isokinetic devices are only designed to measure this load. If the effects due to gravity are added to the external force, the dissimilarity between the algebraic sum of these two forces and the proximal moment is lessened. Therefore, the use of a correction for gravity during isokinetic testing is beneficial.

*Acknowledgement*—This study was sponsored in part by NIH grants AR 26287 and CA 23751 and The Bush Foundation.

## REFERENCES

Andersson, G. B. and Schultz, A. B. (1979) Transmission of moments across the elbow joint and the lumbar spine. *J. Biomechanics* **12**, 747–755.

Andriacchi, T. P., Andersson, G. B. J., Fermier, R. W., Stern,

D. and Galante, J. O. (1980) A study of lower limb mechanics during stair climbing. *J. Bone Jt Surg.* **62-A,** 749–757.

Boccardi, S., Pedotti, A., Rodano, R. and Santambrogio, C. G. (1981) Evaluation of muscular moments at the lower limb joints by an on-line processing of kinematic data and ground reaction. *J. Biomechanics* **14,** 35–45.

Bresler, B. and Frankel, J. P. (1950) The forces and moments in the leg during level walking. *Trans. ASME* **72,** 27–36.

Cavanagh, P. R. and Gregor, R. J. (1975) Knee joint torque during the swing phase of normal treadmill walking. *J. Biomechanics* **8,** 337–344.

Chao, E. Y. (1980) Justification of triaxial goniometer for the measurement of joint rotation. *J. Biomechanics* **13,** 989–1006.

Chao, E. Y. and Hoffman, R. R. (1978) Instrumented measurement of human joint motion. *Instru. Soc. Am. Trans.* **17,** 13–19.

Clauser, C. E., McConville, J. T. and Young, J. W. (1969) Weight, volume, and center of mass of segments of the human body, AMRL-TR-69-70 (AD 710 622). Aerospace Medical Research Laboratory, Wright-Patterson Air Force Base, OH.

Craven, P. and Wahba, G. (1979) Smoothing noisy data with spline functions. *Numerische Mathematik* **31,** 377.

Ericson, M. O. and Nisell, R. (1986) Tibiofemoral joint forces during ergometer cycling. *Am. J. Sports Med.* **14**(4), 285–290.

Gransberg, L. and Knuttson, E. (1983) Determination of dynamic muscle strength in man with acceleration controlled isokinetic movements. *Acta Physiol. Scand.* **119,** 317–320.

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

Herzog, W. (1988) The relationship between the resultant moments at a joint and the moments measured by an isokinetic dynamometer. *J. Biomechanics* **21**(1), 5–12.

Johnson, J. and Siegel, D. (1978) Reliability of an isokinetic movement of the knee extensors. *Res. Q* **49,** 88–90.

Kelley, D. L., Dainis, A. and Wood, G. K. (1976) Mechanics and muscular dynamics of rising from a seated position. In *Biomechanics*, V-B (Edited by Komi, P. V.) pp. 127–134. University Park Press, Baltimore.

Knapik, J. J., Wright, J. D., Mawdsley, R. N. and Braun, J. (1983) Isometric, isotonic, and isokinetic torque variations in four muscle groups through a range of joint motion. *Phys. Ther.* **63,** 938–947.

Lander, J. E., Bates, B. T., Sawhill, J. A. and Hamill, J. (1985) A comparison between free-weight and isokinetic bench pressing. *Med. Sci. Sports Ex.* **17,** 344–353.

Lewis, J. K., Lew, W. D. and Zimmerman, J. R. (1980) A nonhomogeneous anthropometric scaling method based on finite element principles. *J. Biomechanics* **13,** 815–824.

Lindahl, O., Movin, A. and Ringqvist, I. (1969) Knee extension. *Acta Orthop. Scand.* **40,** 79–85.

McConville, J. T., Churchill, T. D., Kaleps, I., Clauser, C. E. and Cuzzi, J. (1980) Anthropometric relationships of body and body segment moments of inertia. Technical Report AFAMRL-TR-80-119, Air Force Aerospace Medical Research Laboratory, Aerospace Medical Division, Air Force Systems Command, Wright-Patterson Air Force Base, OH.

Morrison, J. B. (1968) Bioengineering analysis of force actions transmitted by the knee joint. *Biomed. Engr.* **3,** 164–170.

Nisell, R. (1985) Mechanics of the knee: a study of joint and muscle load with clinical applications. *Acta Orthop. Scand.* (Suppl.) **216,** 42–56.

Osternig, L. R., Hamill, J., Sawhill, J. A. and Bates, B. T. (1983a) Influence of torque and limb speed on power production in isokinetic exercise. *Am. J. Phys. Med.* **62,** 163–171.

Osternig, L. R., Sawhill, J. A., Bates, B. T. and Hamill, J. (1983b) Function of limb speed on torque patterns of antagonist muscles. In *Biomechanics*, Vol. VIII-A, Human Kinetics Publishers, Champaigne, IL.

Perrine, J. J. and Edgerton, V. R. (1978) Muscle force–velocity and power–velocity relationships under isokinetic loading. *Med. Sci. Sports Exer.* **10,** 159–166.

Radcliffe, C. W. (1962) The biomechanics of below knee prostheses in normal level bipedal walking. *Artificial Limbs* **6,** 16.

Radin, E. L. and Paul, I. L. (1972) A consolidated concept of joint lubrication. *J. Bone. Jt Surg.* **54-A,** 607–616.

Sapega, A. A., Nicholas, J. A., Sokolow, P. and Saraniti, A. (1982) The nature of torque overshoot in Cybex isokinetic dynamometry. *Med. Sci. Sports Exer.* **14(5),** 368–375.

Sawhill, J. A., Bates, B. T., Osternig, L. R. and Hamill J. (1982) The variability of isokinetic measures. *Med. Sci. Sports Exer.* **14,** 177.

Schüldt, K., Ekholm, J., Németh, B., Arborelius, U. P. and Harms-Ringdahl, K. (1983) Knee load and muscle activity during exercises in rising. *Scand. J. Rehab. Med.* (Supple.) **9,** 174–188.

Smidt, G. L. (1973) Biomechanical analysis of knee flexion and extension. *J. Biomechanics* **6,** 79–92.

Thorstenson, A., Grimby, G. and Karlsson, J. (1976) Force–velocity relations and fiber composition in human knee extensor muscles. *J. Appl. Physiol.* **40,** 12–16.

Wahrenberg, H., Lindbeck, L. and Eckholm, J. (1978a) Dynamic loading in the human knee joint during voluntary active impact to the lower leg. *Scand. J. Rehab. Med.* **10,** 93–98.

Wahrenberg, H., Lindbeck, L. and Eckholm, J. (1978b) Knee muscular moment, tendon tension force and EMG during a vigorous movement in man. *Scand. J. Rehab. Med.* **10,** 99–106.

White, A. A. and Panjabi, M. M. (1978) *Clinical Biomechanics of the Spine.* J. B. Lippincott, Philadelphia.

Winter, D. A. (1980) Overall principle of lower limb support during stance phase of gait. *J. Biomechanics* **13,** 923–927.

Winter, D. A. (1983) Moments of force and mechanical power in jogging. *J. Biomechanics* **16,** 91–97.

Winter, D. A., Wells, R. P. and Orr, G. W. (1981) Errors in the use of isokinetic dynamometers. *J. Appl. Physiol.* **46,** 397–408.

Woltring, H. J. (1985) On optimal smoothing and derivative estimation from noisy displacement data in biomechanics. *Hum. Mov. Sci.* **4,** 229–245.

Woltring, H. J. (1986) A Fortran package for generalized, cross-validatory spline smoothing and differentiation. *Adv. Engr. Software* **8**(2), 104–113.

Zernicke, R. F., Garhammer, J. and Jobe, F. W. (1977) Human patellar-tendon rupture. *J. Bone Jt Surg.* **59-A,** 179–183.