The Study of Kinematic Transients in Locomotion Using the Integrated Kinematic Sensor

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***Abstract*—A system based on the integrated kinematic sensor (IKS) was used to study the three-dimensional (3-D) kinematics of human lower limb during walking and running. The linear displacement, angular velocity, and linear acceleration of the foot, shank, and thigh segments were directly measured using three IKS's. The results clearly showed the heel strike impact in both walking and running, illustrating the high frequency components that exist in those activities. This paper illustrates the limitations of standard position measurements to capture transients associated with phase transitions, not only in acceleration estimates, but also in the determination of segmental angular velocities. An error analysis was conducted to determine the relative contribution of the accelerometer and the angular rate sensor to the determination of the segmental center of mass (COM) acceleration. The results suggest that in practical kinesiological applications, adding either an accelerometer or an angular rate sensor can remarkably increase the accuracy of segmental COM acceleration estimates.**

I. Introduction

Transition between swing and stance phases during locomotion is one of the fundamental features of movement. Basically, there are two kinds of transitions: heel strike and toe off. While toe off refers to the transition from stance to swing, during which a certain amount of push-off force is generated in order to achieve the set swing speed, heel strike refers to the transition from swing to stance, during which an impact force is loaded to the lower limb. The magnitude of the impact depends on, among other things, the speed of the locomotion (e.g., walking versus running), the type of activity (e.g., walking versus high jump landing), the footwear (e.g., barefoot versus sneaker), and the walking surface (e.g., concrete versus grass).

Acute injuries in the lower limb occur mainly at the transition from swing to stance phase [1]. It has been demonstrated that violent stress imposed to the quadriceps to gain control after landing a jump can result in an anterior curciate ligament (ACL) rupture [2]; a lateral impact force applied to the knee can cause a medial collateral ligament (MCL) and ACL injury [3]; and a shear impact force at the hip joint upon falling can easily exceed the femoral fracture threshold (about 2 kN) and lead to hip fracture [4], [5].

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Because the kinematics at these transitions have higher frequency components than during other portions of the movement, the traditional biomechanical approaches in quantifying body kinematics and kinetics, such as using an image-based motion analysis system, are not satisfactory. The difficulty with the direct measurement of body movement using an image-based motion analysis system is associated with the necessity of low-pass filtering for the time derivatives of the position measurement. The cut-off frequencies of the low-pass filter applied to human gait studies have typically been around 5–7 Hz [6]. As a result, the presented kinematics as well as the kinetics of human locomotion lack high frequency components [7]–[9].

A different approach for studying the body segmental kinematics is based on the direct measurement of the body segmental acceleration using accelerometers [10], [11]. For example, in a study of the effect of footwear on kinematics of locomotion, Light *et al.* [12] attached linear accelerometers on the leg and revealed the longitudinal acceleration impact upon heel strike. However, due to the nonlinear relation between the linear and angular kinematic variables, multiple noncollinear accelerometers (between 5 and 12) attached to a single segment have been used to resolve its complete kinematics [10], [13].

This paper reports on the use of the IKS for studying kinematic transients during high-impact activities such as heel-strike. The integrated approach combines position, linear acceleration, and angular velocity measurements with the six-degrees-of-freedom analysis of rigid body motion. This approach increases the frequency band of the measured kinematic variables, and therefore reveals some of the high frequency phenomena not reported earlier. This paper also examines the relative contributions of the different components of the sensor, focusing on the importance of angular velocity or linear acceleration measurements in the determination of the linear acceleration at the segmental COM.

II. Methodology

A. Segmental Kinematics Measurements by IKS

The determination of the segmental kinematics based on the output from the IKS was described by Wu and Ladin [14] and is therefore only summarized here. The output of an accelerometer—$\vec{a}_p^{(m)}$ (the superscript m stands for *measured*) attached to an arbitrary point p on a rigid body (assuming that a body segment is rigid) is given by the sum of two terms—the

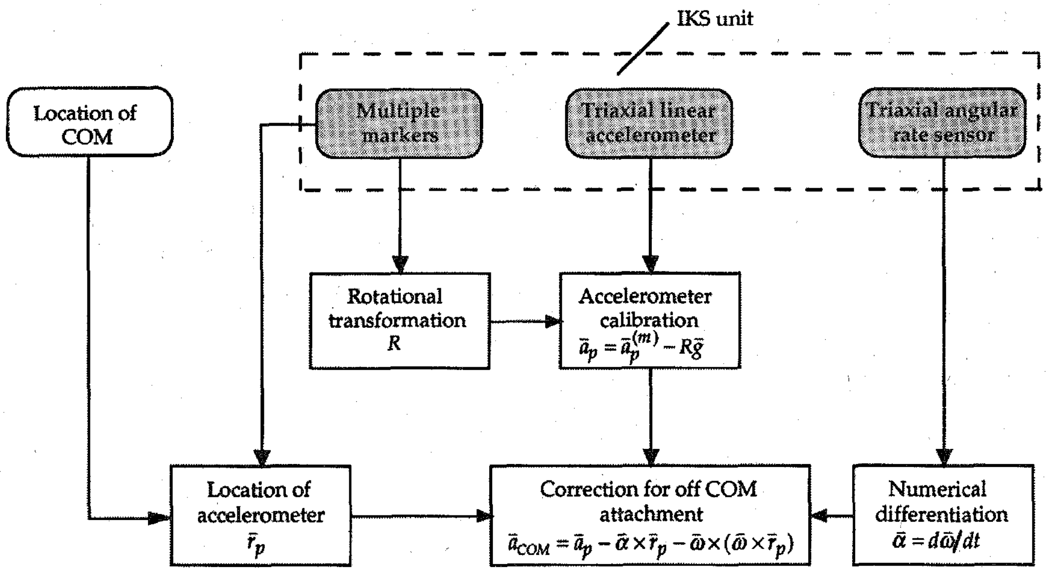


Fig. 1. Schematic diagram of data processing procedure for an IKS.

kinematic acceleration $\vec{a}_p$ at point p due to the motion and the gravitational acceleration $\vec{g}$

$$\vec{a}_p^{(m)} = \vec{a}_p + R_{BL}\vec{g} \tag{1}$$

where R_{BL} is the rotational transformation matrix from the laboratory coordinate system (LCS) to the body-fixed coordinate system (BCS). If the COM of the rigid body is chosen as the origin of the BCS, the relation between the acceleration at COM $\vec{a}_{\text{COM}}$ and $\vec{a}_p^{(m)}$ is given by the following equation:

$$\vec{a}_{\text{COM}} = \vec{a}_p^{(m)} - R_{BL}\vec{g} - \vec{\alpha} \times \vec{r}_p - \vec{\omega} \times (\vec{r}_p) \tag{2}$$

where $\vec{\omega}$ and $\vec{\alpha}$ are the angular velocity and acceleration of the rigid body, respectively, $\vec{r}_p$ is the distance of point p to the COM.

The IKS combines direct measurements of segmental position, linear acceleration, and angular velocity to allow accurate estimation of COM acceleration. The detailed procedure for data processing is depicted in Fig. 1. The direct measurement of linear displacements of multiple markers on a rigid body has two purposes. First, it is used to determine the spatial orientation of the body and the rotational transformation matrix R_{BL}. This rotational transformation matrix is used to dynamically calibrate the linear accelerometer because of its sensitivity to the field of gravity [15]. Second, it is used to determine the relative position of the accelerometer to a reference point. Given that the relative positions of the COM and the accelerometer to a reference point are known, the relative position of the accelerometer to the COM $(\vec{r}_p)$ can be determined. In addition, the direct measurement of angular velocity of the body segment is first used to determine the angular acceleration by a backward, two point numerical differentiation method. Both angular velocity and acceleration are then used to correct for the off COM attachment. The three orthogonal angular velocity components of a rigid body are measured by one triaxial angular rate sensor.

B. Error Analysis

The relative importance of different components in the IKS was examined by error analysis in the calculation of the acceleration at the COM of the segment as given in (2). Since each kinematic variable in (2) is either directly measured or calculated (except for $\vec{r}_p$ and $\vec{g}$), each variable is assumed to have a noise component $\vec{n}$. That is:

$$\begin{aligned}\hat{\vec{a}}_p^{(m)} &= \vec{a}_p^{(m)} + \vec{n}_a\\ \hat{\vec{\omega}} &= \vec{\omega} + \vec{n}_\omega\\ \hat{\vec{\alpha}} &= \vec{\alpha} + \vec{n}_\alpha\\ \hat{R}_{BL} &= R_{BL} + n_R.\end{aligned} \tag{3}$$

Replacing these variables into (2), we have

$$\begin{aligned}\hat{\vec{a}}_{\text{COM}} &= \hat{\vec{a}}_p^{(m)} - \hat{R}_{BL}\vec{g} - \hat{\vec{\alpha}} \times \vec{r}_p - \hat{\vec{\omega}} \times (\hat{\vec{\omega}} \times \vec{r}_p)\\ &= \vec{a}_p^{(m)} - R_{BL}\vec{g} - \vec{\alpha} \times \vec{r}_p - \vec{\omega} \times (\vec{\omega} \times \vec{r}_p)\\ &\quad + \vec{n}_a - n_R\vec{g} - \vec{n}_\alpha \times \vec{r}_p - \vec{\omega} \times (\vec{n}_\omega \times \vec{r}_p)\\ &\quad - \vec{n}_\omega \times (\vec{\omega} \times \vec{r}_p) - \vec{n}_\omega \times (\vec{n}_\omega \times \vec{r}_p)\\ &= \vec{a}_{\text{COM}} + \Delta\vec{a}_{\text{COM}}\end{aligned} \tag{4}$$

where

$$\begin{aligned}\Delta\vec{a}_{\text{COM}} &= \vec{n}_a - n_R\vec{g} - \vec{n}_\alpha \times \vec{r}_p - \vec{\omega} \times (\vec{n}_\omega \times \vec{r}_p)\\ &\quad - \vec{n}_\omega \times (\vec{\omega} \times \vec{r}_p) - \vec{n}_\omega \times (\vec{n}_\omega \times \vec{r}_p)\end{aligned} \tag{5}$$

is the error in the calculation of COM acceleration. In the worst situation, that is, when all the resulting vectors in (5) are in the same direction, the maximum error can be determined as

$$\begin{aligned}|\Delta\vec{a}_{\text{COM}}|_{\max} \le{}& |\vec{n}_a| + |n_R||\vec{g}| + (|\vec{n}_\alpha|\\ &+ 2|\vec{\omega}||\vec{n}_\omega| + |\vec{n}_\omega|^2)|\vec{r}_p|.\end{aligned} \tag{6}$$

This maximum error was estimated for the following cases in which the noise for direct measurement is assumed to be white with a magnitude of N.

(a) (b) (c) (d)

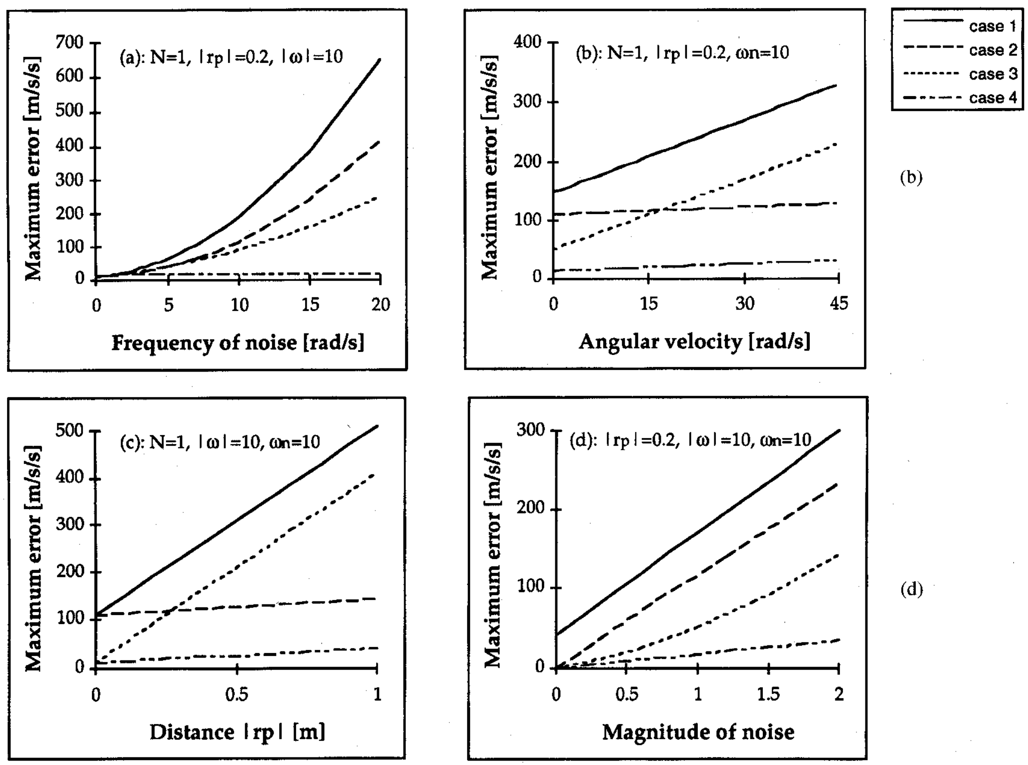


Fig. 2. Maximum errors in the calculation of COM acceleration in four cases in relation to (a) frequency of the noise; (b) magnitude of segmental angular velocity; (c) distance of the measurement site to the COM; and (d) magnitude of the noise.

Case 1: Direct measurement of position only—velocity and accelerations are obtained through first or second order differentiation based on position. The noises are

$$\begin{aligned} |n_R| &= N \\ |\vec{n}_\omega| &= N\omega_n \\ |\vec{n}_a| &= |\vec{n}_\alpha| = N\omega_n^2 \end{aligned} \tag{7}$$

where ω_n is the frequency of the noise.

Case 2: Direct measurement of position and angular velocity—the accelerations are obtained by differentiation. The noises are

$$\begin{aligned} |n_R| &= |\vec{n}_\omega| = N \\ |\vec{n}_\alpha| &= N\omega_n \\ |\vec{n}_a| &= N\omega_n^2. \end{aligned} \tag{8}$$

Case 3: Direct measurement of position and linear acceleration—the angular velocity and acceleration are obtained by differentiation. The noises are

$$\begin{aligned} |n_R| &= |\vec{n}_a| = N \\ |\vec{n}_\omega| &= N\omega_n \\ |\vec{n}_\alpha| &= N\omega_n^2. \end{aligned} \tag{9}$$

Case 4: Direct measurement of position, angular velocity, and linear acceleration—the angular acceleration is obtained by differentiation. The noises are

$$\begin{aligned} |n_R| &= |\vec{n}_\omega| = |\vec{n}_a| = N \\ |\vec{n}_\alpha| &= N\omega_n. \end{aligned} \tag{10}$$

Substituting (6) with (7) through (10) accordingly, the maximum error for each of the four cases is derived as following.

Case 1:

$$\begin{aligned} |\Delta\vec{a}_{\mathrm{COM}}|_{\max} \le {} & |\vec{g}|N + 2|\vec{\omega}||\vec{r}_p|N\omega_n \\ & + (1 + |\vec{r}_p| + N|\vec{r}_p|)N\omega_n^2 \end{aligned} \tag{11}$$

Case 2:

$$\begin{aligned} |\Delta\vec{a}_{\mathrm{COM}}|_{\max} \le {} & [|\vec{g}| + (2|\vec{\omega}| + N)|\vec{r}_p|]N \\ & + |\vec{r}_p|N\omega_n + N\omega_n^2 \end{aligned} \tag{12}$$

Case 3:

$$\begin{aligned} |\Delta\vec{a}_{\mathrm{COM}}|_{\max} \le {} & (1 + |\vec{g}|)N + 2|\vec{\omega}||\vec{r}_p|N\omega_n \\ & + (1 + N)|\vec{r}_p|N\omega_n^2 \end{aligned} \tag{13}$$

Case 4:

$$|\Delta\vec{a}_{\mathrm{COM}}|_{\max} \le [1 + |\vec{g}| + (2|\vec{\omega}| + N)|\vec{r}_p|]N + |\vec{r}_p|N\omega_n. \tag{14}$$

Clearly, these errors depend on the magnitudes of angular velocity $|\vec{\omega}|$, distance from point p to COM $|\vec{r}_p|$, the noise level N, and the noise frequency ω_n. A graphical representation of these errors as a function of these variables is shown in Fig. 2. In general, the IKS approach (Case 4) presents the least error and the direct position measurement (Case 1) presents the greatest error for all the independent variables. Adding one more direct measurement of either angular velocity or linear acceleration to the position measurement can remarkably

TABLE I
MEAN AND STANDARD DEVIATION OF THE SPEED, ANGULAR VELOCITY, AND LINEAR ACCELERATION FOR THE WALKING AND RUNNING EXPERIMENTS

Speed/ Segment	Axes	Walking		Running	
		ω [rad/s]	a [m/s²]	ω [rad/s]	a [m/s²]
Foot	M-L	11.3 (0.2)	48.2 (8.9)	48.0 (20.9)	152 (34.0)
	A-P	9.7 (0.9)	48.4 (8.8)	68.5 (19.0)	284.3 (41.7)
	P-D	6.5 (0.6)	26.3 (1.4)	36.6 (12.1)	252.0 (34.0)
Shank	M-L	7.0 (0.3)	11.6 (1.7)	12.2 (0.3)	38.3 (6.5)
	A-P	4.1 (0.2)	16.4 (2.7)	5.2 (0.3)	75.8 (18.4)
	P-D	4.5 (0.2)	19.7 (0.7)	7.5 (1.1)	67.8 (7.9)
Thigh	M-L	3.6 (0.1)	15.0 (1.9)	8.0 (0.3)	39.6 (5.6)
	A-P	1.7 (0.2)	24.6 (5.8)	2.8 (0.4)	76.9 (4.8)
	P-D	2.2 (0.5	11.2 (1.6)	5.9 (0.1)	88.2 (7.2)

reduce the maximum error. However, the relative importance of these two additional measurements in the error reduction depends heavily on those independent variables. For example, adding the direct measurement of linear acceleration (Case 3) is more significant in reducing the error across high frequency range than adding the direct measurement of angular velocity (Case 2), whereas it is the opposite when the body moves at high speed (i.e., $|\vec{\omega}|$is large) or the direct measurement site is far away from the COM (i.e., $|\vec{r}_p|$ is large).

C. Experimental Evaluation

The kinematics of the foot, shank, and thigh during walking and running was investigated on three healthy young male subjects (weight: 74.5/70.0/73.5 Kgf and height: 181.6/176.5/167.6 m). Three IKS's (weight about 0.1 kg each) were attached to the dorsal surface of the right foot and to the frontal surface of the right shank and thigh by padded plastic casts that were molded to fit the shapes of the lower limb segments. This mounting helped to evenly distribute the load of the IKS to the body segment. The maximum ranges for the accelerometers (Entran Devices, Fairfield, NJ) were 1000 g, 100 g, and 10 g for the foot, shank, and thigh, respectively, and the ranges for the angular rate sensors (Watson Industries, Inc., Eau Claire, WI) were 1500 deg/s for all the body segments. For static calibration, markers were attached to the landmarks of the body segments; that is, lateral malleolus, tibial tubercle and greater trochanter, so that the relative locations of the IKS's with respect to these landmarks were determined. In addition, the anthropometric parameters of these body segments were manually measured and the COM locations of the foot, shank, and thigh segments with respect to these landmarks were calculated based on the regression equations developed by Zatsiorsky and Seluyanov [16]. Therefore, the relative position of each of the IKS's to the COM of the corresponding body segment can be calculated.

The walking and running tasks were performed on a 30 m runway. Two cameras from the WATSMART optoelectronic system were placed about 1.5 m apart in front of the runway to cover a viewing volume of 1 m × 2 m × 2 m (in the medial-lateral, anterior-posterior, and vertical directions, respectively). A timer was used to monitor the average speed over a 3.66 m distance.

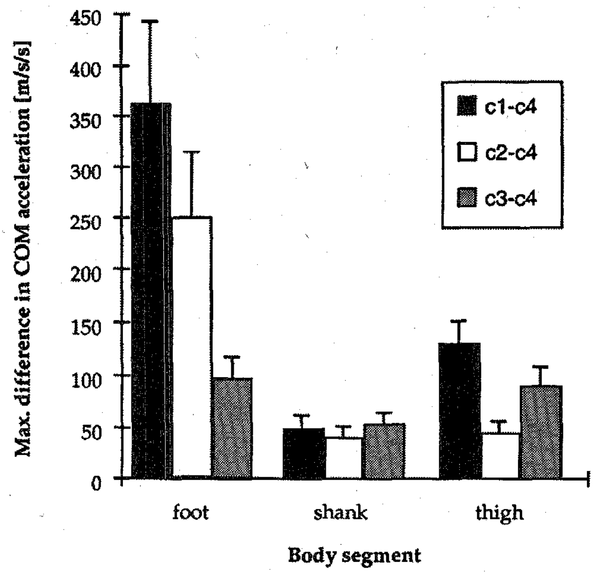
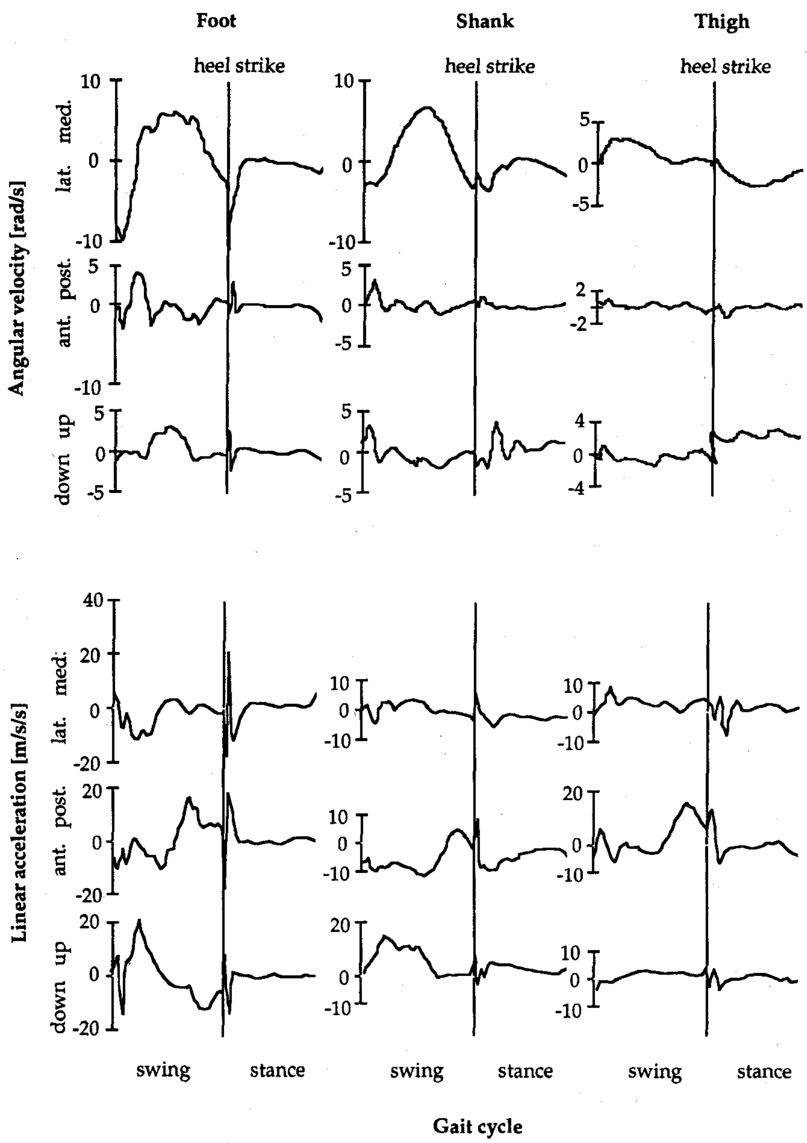


Fig. 3. Means and standard deviations of the maximum differences in the COM acceleration of the foot, shank, and thigh as calculated in cases 1 through 3 against case 4.

Once the IKS's were attached to the subjects, a 10–15 min warm-up period was allowed while they walked freely in the lab in order to get used to the instrumentation. Before each task, the subjects were allowed to practice a few times to settle down the speed and the pace so that one complete gait cycle would be inside the camera's viewing volume. They were instructed to either slow down or speed up at the end of each trial if the speed of the previous trial was not within the pre-set speed range. For each task, multiple trials were collected, but only those trials that were at the required speed were accepted. All the tests were done with bare feet.

All the kinematic variables were low-pass filtered at 100 Hz with three pole active filters, and digitized at 200 Hz by two 12 bit A/D converters synchronized at the beginning of the data collection. The data processing was done on a VAX 11/750 computer. The segmental COM accelerations were calculated based on (2) by utilizing the angular velocity and/or linear acceleration values that were obtained either by direct measurement or by numerical differentiation (the backward



(a)

Fig. 4. IKS measurements of angular velocity and linear acceleration of the foot, shank, and thigh of one subject around the heel strike of walking (a).

difference method) of position measurement. A digital low-pass filter was applied to the kinematic variables that were obtained by numerical differentiation. The filter was a third order Butterworth applied forward and backward in time to eliminate the phase shift.

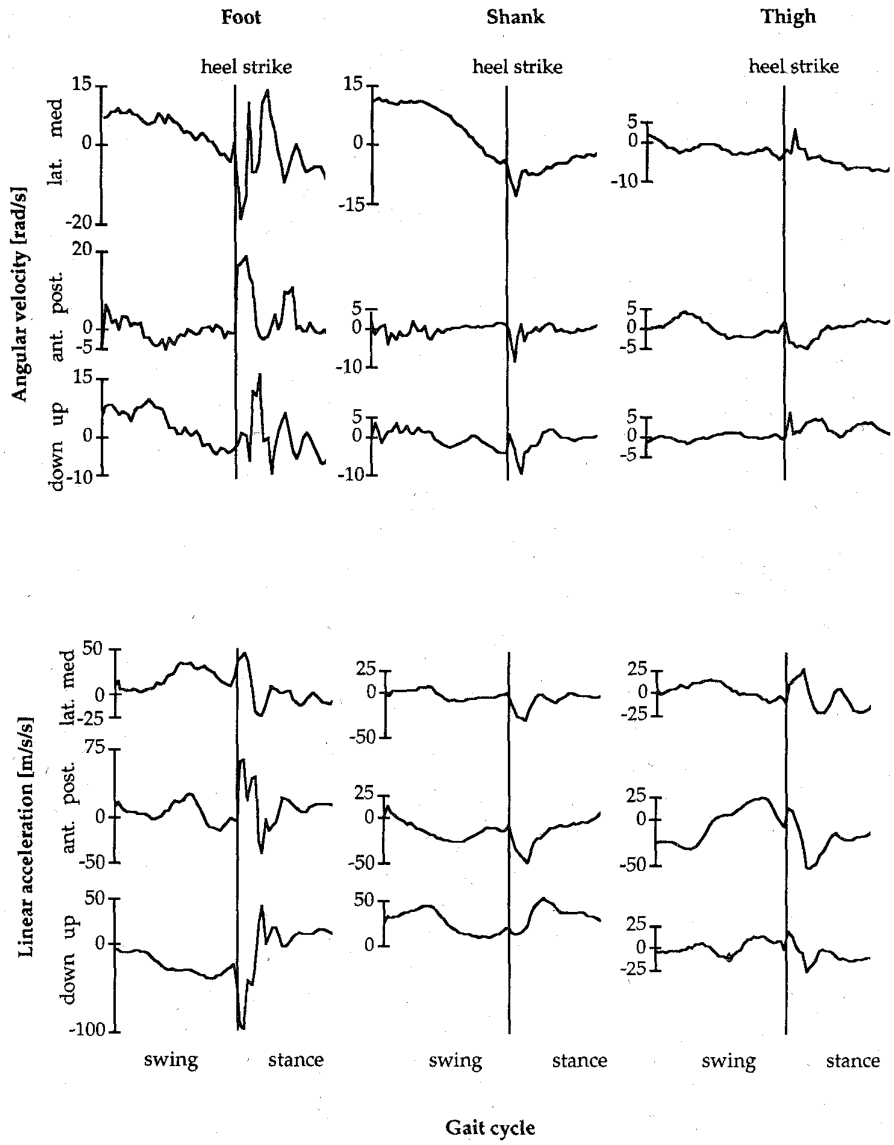
III. Results

A. Repeatability

The average speed for ten walking trials was 1.34 m/s with standard deviation of 0.04 m/s, and the average speed for nine running trials was 3.93 m/s with variation of 0.12 m/s. The within subject variations of the angular velocity as well as linear acceleration measurements of the shank and thigh were small for both walking and running. The mean and the standard deviations of the maximum peak values of the angular velocity and linear acceleration of three body segments over one complete gait cycle are summarized in Table I. Although these measurements varied more on the foot than on the shank and thigh, the variation was less than 1 rad/s for angular velocity and less than 1 g for linear acceleration during walking.

B. Acceleration Error in the Lower Limb

The maximum errors in the segmental COM acceleration, i.e., the difference between the time trajectory of case 4 and those estimated by either cases 1, 2, or 3 are illustrated in Fig. 3 for the foot, shank, and thigh segments. These errors occur during the swing-stance transition. Consistent with the findings shown in Fig. 2, the errors in case 1 were the largest, whereas the errors in cases 2 and 3 varied depending on the specific body segment. For example, case 3 showed much less error than case 2 for estimating the foot acceleration,



(b)

Fig. 4 *(Continued.)* IKS measurements of angular velocity and linear acceleration of the foot, shank, and thigh of one subject around the heel strike of running (b).

suggesting that the critical element in estimating the foot acceleration is the linear accelerometer. In contrast, case 2 was better than case 3 in estimating shank and thigh accelerations, suggesting that the angular velocity is the critical variable for these segments. This could be partially due to the difference in $\vec{r}_p$ and in the magnitude of angular velocity of three body segments. Nevertheless, adding either the linear acceleration or the angular velocity measurement could reduce the maximum error at heel strike comparing to the error obtained in case 1.

C. Heel Strike Transient

Because of the advantage of IKS in being able to capture high frequency content in high-impact activities, both angular velocity and linear acceleration of the lower limb, as measured by IKS, at heel strike transient during walking and running were examined and shown in Fig. 4. Clearly, large impulses at heel strike were observed for all the situations, although the peak impulses for the proximal segments (such as shank and thigh) were attenuated due to the shock absorbent effect of the limb structure. In addition, the heel strike impulses increased remarkably as the movements become more impulsive in nature (such as running). Moreover, these impulses occurred not only in the plane of the primary movement, they were observed in other planes as well. The foot acceleration in the medial-lateral direction, for example, was as high as 4 g for walking and more than 10 g for running.

IV. Discussion

Partial kinematic measurements of the angular velocity and linear acceleration of some lower-limb segments have been reported in the past. Morris [10] used multiple accelerometers to measure shank acceleration during the swing phase of

TABLE II
MAXIMUM PEAK ACCELERATIONS REPORTED IN THE LITERATURE

Investigators	Speed	Axes	Foot	Shank	Thigh
This study	1.31m/s (walking)	M-L	4.9g	1.2g	1.5g
		A-P	4.9g	1.6g	2.5g
		P-D	2.6g	2g	1.1g
	3.84m/s (running)	M-L	15g	3.8g	4g
		A-P	28g	7.6g	7.7g
		P-D	25g	6.8g	8.9g
Lafortune, [1991]	1.5m/s (walking)	M-L		1.3g	
		A-P		2g	
		P-D		2.3g	
	3.5m/s (running)	M-L		4.7g	
		A-P		7.3g	
		P-D		3g	
Radin et al., [1991]	1.37m/s	P-D	-	1.6g	1.3g
Gilbert et al., [1984]	0.6s/stance (walking)	A-P	-	2g	3g
		P-D	-	2g	2.5g
Light et al., [1980]	(walking)	P-D	-	6g	-
Moris, [1973]	(walking)	M-L	-	0.5g	-
		A-P	-	1.5g	-
		P-D	-	1g	-

walking, and *estimate* the angular velocities in the coronal and sagittal planes. Gilbert *et al.* [11] used eight uniaxial accelerometers to measure the linear accelerations of the shank and thigh segments in the sagittal plane during the stance phase of walking. Light *et al.* [12] studied the heel strike transients by two accelerometers, one fixed to the tibia and the other attached to the shank as close as possible to the tibia. Radin *et al.* [17] used three single axis accelerometers to measure the linear accelerations of the shank and thigh in the sagittal plane, and *estimate* the angular velocities of those body segments. To the best of our knowledge direct measurements of the angular velocities of the lower-limb segments have not been reported thus far.

The usefulness of adding measurement devices (such as accelerometer or angular rate sensor) has to be weighed considering the benefits of the added information. It appears that adding a single measurement device (either an accelerometer or an angular rate sensor) can remarkably increase the frequency content of the corresponding kinematic variable, the estimated segmental COM acceleration and all the dynamic variables (i.e., joint forces and moments). This would improve the accuracy with which transient phenomena can be studied. The value of such information clearly increases as the overall speed of the underlying activity increases (e.g., running versus walking). However, it appears that a single sensor does not necessarily reduce the error in estimating the segmental COM acceleration, as demonstrated by Figs. 2 and 3. Adding both sensors would therefore improve the estimate of the high frequency components, while reducing the noise level throughout the frequency band.

Kinematic variables directly measured by the IKS showed good repeatability within subjects except for some components characterizing the foot kinematics, suggesting that the differences are due to different walking styles of individual subjects. Most importantly, the transients in both angular velocity and linear acceleration were clearly observed during heel strike. The peaks of these impulses were most distinct at the foot and smallest at the thigh. The directions of the transients were highly repeatable between different trials within individual subjects as well as across subjects during walking and running. Furthermore, the peak values increased greatly from walking to running, while the speed of locomotion increased by a factor of three only. Further studies are needed to evaluate the relation between the magnitude of the impulses to the nature of the injuries associated with physical activities.

The results obtained by the IKS are generally in agreement with those reported by other research groups. Gilbert *et al.* [11], Light *et al.* [12], Radin *et al.* [17] and Lafortune [18] either described or reported distinct heel strike impulses for both the shank and thigh segments in all the observed directions. The quantitative comparison of the acceleration measurements by this study to the above research groups is summarized in Table II, in which the maximum peak values of the reported lower limb accelerations are listed. The comparison is limited to the accelerations of the shank and thigh segments during walking and running, as these are the *only* values that have been reported. In general, the maximum magnitudes of the shank accelerations reported in this study are comparable to others except that the peak magnitudes of the vertical shank acceleration in this study are different from that reported by Light *et al.* [12] in walking and by Lafortune [18] in running. This difference is probably due to the difference in the experimental conditions: Our data was collected while the subject was barefoot, while Light *et al.* and Lafortune had their subjects wearing footwear during testing. It is important to note that the results summarized in Table II reflect a variety of experimental conditions. For example, the exact speeds of the activities described by Light *et al.* [12] and Morris [10], as well as the types of footwear used by Morris are not given. Moreover, although the COM accelerations of the shank and thigh were reported by Gilbert *et al.* [11], the different approaches used for estimating the COM location would introduce errors in the COM acceleration calculation. Therefore, it is difficult to conduct direct quantitative comparisons among different research groups. However, in spite of the above limitations, our results are in agreement with the nature and order of magnitude of the reported accelerations.

It should be admitted that attaching IKS's to the lower limb does add additional weight to the body, which can potentially alter the normal gait pattern. In this study, however, several attempts have been made to reduce this artifact. They include choosing light weighted transducers, evenly distributing the load of the IKS to the body segment through molded casts, and allowing the subjects to warm up before data collection so that they could perform with little interference from the IKS's. As a result, all the subjects rated the instrumentation attachment comfortable. The fact that the results as reported in this study are in agreement with others as listed in Table II suggests that the potential effect of using IKS on gait pattern changes can be limited. Clearly, all the subjects tested in the study are healthy young adults. The applicability of using the IKS approach on other populations such as young children, elderly, or pathological groups, needs to be tested in future studies.

V. Conclusions

The IKS approach was used to study the kinematic transients of the human lower limb during walking and running. It combines three kinematic measurements of rigid body motion: the position, angular velocity, and linear acceleration. Segmental COM accelerations were compared to the literature and to those estimated by differentiation approach. It is demonstrated that the angular velocity and linear acceleration *measurements* play a dominant role in increasing the frequency range of the estimated segmental COM accelerations. The IKS approach eliminates the low-pass filtering required by the differentiation approach, thereby substantially increasing the frequency range of the kinematic observations during transients and during high speed activities. In addition, the IKS approach eliminates the problems existing in the current accelerometry approach, so that neither the multiaccelerometer scheme nor time integration is necessary.

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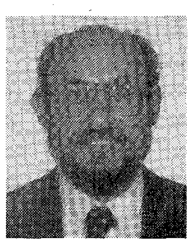

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