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Validation of an Inertial Sensor System for Quantifying Knee Function

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VALIDATION OF AN INERTIAL SENSOR SYSTEM FOR QUANTIFYING KNEE FUNCTION

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ABSTRACT

Gait analysis has become a useful tool for clinicians in evaluating the progression of pathologies through functional analysis. The high cost and dedicated laboratories associated with the traditional camera-based motion analysis systems present the need for an alternative system. Direct measurement of kinetic parameters using inertial sensors (gyroscopes and accelerometers), in place of indirect calculations from position data obtained using cameras, has been shown effective in resolving important gait parameters.

In order to directly compare gait parameters obtained using inertial sensors and a camera system, data was simultaneously collected from both systems for seven test subjects during normal gait. Three uni-axial gyroscopes and one tri-axial accelerometer were mounted on each subject's right leg, as well as the reflective markers needed for the camera-based system. Knee flexion angle, angular velocities, and linear and angular accelerations were compared between the two systems.

The similarities between the two methods validate the accuracy of the inertial sensor system with respect to the currently accepted camera-based method for some parameters. The errors found when comparing the two systems can be minimized by altering the number of sensitive axes of the

sensors, as well as improving the accuracy of their placement. Such an inertial sensor system may provide an alternative that is suitable for use in a clinical setting.

DEDICATION

I would like to dedicate this manuscript to my family for all the support that they have given me throughout graduate school. In particular my Mom, for having to listen to my phone calls when I had problems along the way, and also to my grandparents for always giving me the encouragement to work through the barriers I encountered and towards completion when I especially needed it.

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TABLE OF CONTENTS

Table of Contents (Continued)

LIST OF TABLES

LIST OF FIGURES

CHAPTER 1

SENSOR STUDY

Introduction

In recent years, human motion analysis has been transformed from a research tool [1] to clinical evaluation method used in a variety of applications [2, 3]. Current motion analysis systems, such as video-capture systems, infrared camera systems and electromagnetic motion measuring systems involve the use of multiple cameras mounted within large dedicated laboratory space. The necessary equipment and space make these systems expensive, and limit the scope of their use to a laboratory setting [4, 5]. The cost of equipment necessary to start a camera-based motion laboratory averages $300,000, and because of the high cost of data capture and analysis, the clinical use of gait analysis is limited [6]. For the implementation of motion analysis in a clinical environment, a small portable sensor system is an ideal tool for quantifying the gait characteristics of various disorders. The utilization of accelerometers and gyroscopes, which provide a less costly but still accurate alternative, has been examined as an option that is portable and does not limit the range of motion of the subject to within the capture region of cameras [7]. Studies have shown promising results for systems designed to collect data for long time periods outside of a controlled environment, and for different activities of daily living such as rising from a chair [8] or going up and down stairs [9]. Such a portable system would allow the collection of biomechanical data for patients in a wide variety of settings, and at a lower cost per test [5, 10, 11].

Advantages of using inertial sensors (accelerometers and gyroscopes) include their small size, low cost compared to camera-based systems, and their portability. They provide direct measurement of the accelerations and angular velocities of a body segment, respectively. This limits loss of data due to filtering and the derivation of accelerations and velocities from position data as with camera-based systems. Angular orientations and relative angles of body segments can be calculated through strap-down integration of angular velocity data.

The purpose of this study is to validate the accuracy of an inertial sensor system for motion analysis through a direct comparison of gait parameters with those obtained simultaneously from a standard camera-based motion analysis system. This study will explore the use of inertial sensors as an alternative method for camera-based motion analysis.

Methods

In order to validate the inertial sensor system, test subjects were instrumented with both the inertial sensors and the reflective markers for a camera-based system. Data were simultaneously collected with both systems, and kinetic and kinematic parameters were calculated.

Instrumentation

A set of four inertial sensors were wired to a laptop-based data acquisition system (National Instruments®, Austin, TX): a tri-axial accelerometer (Model EGA-3 Entran Devices Inc, Fairfield, NJ; +/- 10g range) and three single axis rate gyroscopes (ADXRS300, Analog Devices, Inc., Wilmington, MA; +/- 300 deg/sec range).

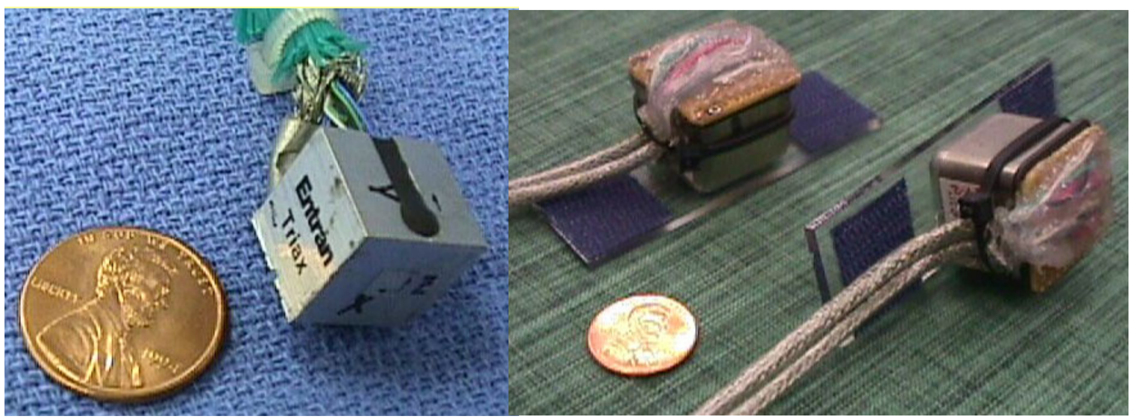


Figure 1. The picture on the left is of the tri-axial accelerometer used for the study and on the right are the gyroscopes that were placed on the shank and thigh.

The gyroscopes were mounted onto thin rigid plastic strips, in order to allow for secure placement onto the leg. One of the gyroscopes was placed on the anterior aspect of the shank with its sensitive axis aligned with the long axis of the shank. Two gyroscopes were placed in the sagittal plane on the lateral side of both the shank and thigh directly against the skin. The output of a gyroscope is not sensitive to its position along its sensitive axis on a rigid body, so the gyroscopes were positioned to minimize the soft-tissue interference. Elasticized Velcro® bands were used to hold the sensors in place. After the sensors were properly aligned and the Velcro straps were secured, the sensors were further secured to the leg with Coban® (3M, St. Paul, MN) self-adherent wraps to ensure they did not move. The tri-axial accelerometer was placed adjacent to the gyroscope on the anterior aspect of the shank, as close to the center of mass of the shank as possible. It was secured with Velcro® straps and Coban® along with the gyroscopes.

Accelerometer Calibration

Since the three axes of the accelerometer could not be aligned to coincide with the shank axis system, it was necessary to transform the axis system of the accelerometer to that of the shank.

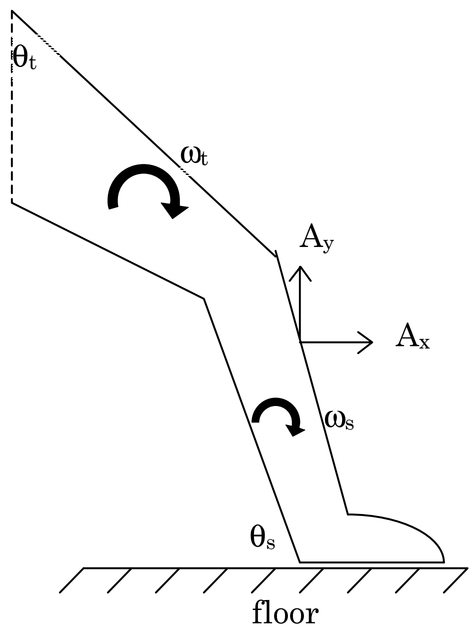


Figure 2. The schematic shows a representation of shank angular velocity (ω_s), thigh angular velocity (ω_t), shank orientation (θ_s) and thigh orientation (θ_t). Ax and Ay represent the orientation and location of two of the linear accelerations measured by the accelerometer.

The orientation of each accelerometer axis with respect to the shank was determined by recording accelerometer output while the subject's leg was positioned so that each shank axis was sequentially aligned with the global coordinate system. This was repeated for each of the 3 shank axes.

Gyroscope Calibration

A common problem with gyroscopes is the drift that can occur over long periods of time and during turns. In previous studies it has been shown that drift can be corrected for each trial by using initial and terminal values measured by the gyroscope to determine the rate of drift and use it to adjust the data when the two values are known by another method of measurement or designing the trial to begin and end at rest where angular velocity is 0°/sec [12]. The data collected in this study were of short duration trials, with the subjects starting and ending at rest.

Simultaneous Data Collection

Data was collected in the Motion Analysis Laboratory at Shriner's Hospital for Children in Greenville, SC using a Vicon® 512 system with 12 M-cameras (Vicon, CA) Each subject was first instrumented with the gyroscopes and accelerometer. After sensor calibration, the reflective markers were applied to the appropriate anatomical landmarks for lower limb gait analysis. Static trials were conducted to determine the center of rotation for the knee for the camera-based system. Each subject was then positioned approximately two strides from a force plate, and walked across the force plate in three to four strides. Data was collected from both systems for two to four seconds before and after motion. To synchronize the data collection between the two systems, a sawtooth signal from a function generator was acquired simultaneously, and waveform peaks were used to align the data temporally.

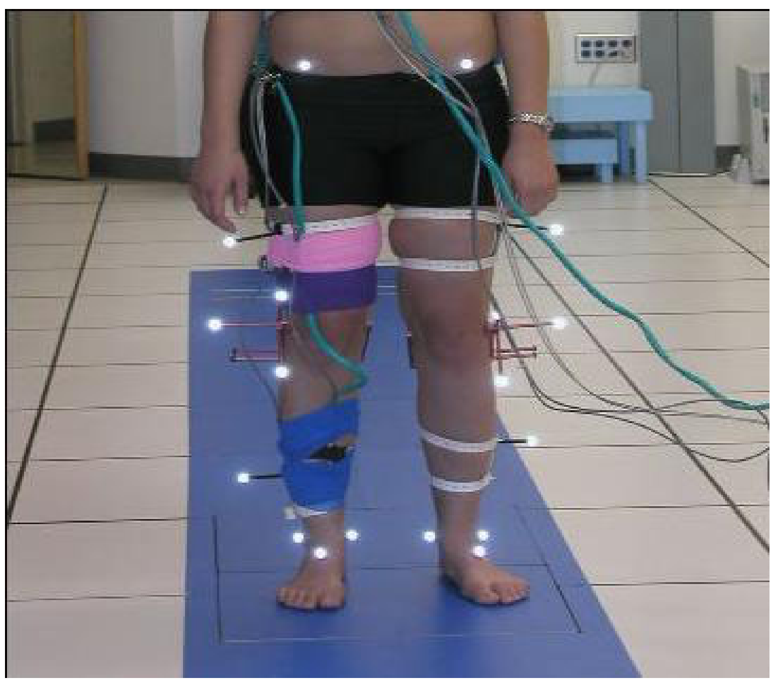


Figure 3. The inertial sensors are wrapped in Coban® and the reflective markers are placed in their proper locations. The patient then walked across the force plates with both systems simultaneously recording data.

Subject Selection

Three or four trials were conducted for seven (5 males and 2 females) normal test subjects with a mean age of 25 ± 9 years with a range of 18-45 years, an average height of 5’8” ± 4” with a range of 5’1”-6’0”, an average weight of 174 ± 49 lbs. with a range of 100-210 lbs, and an average body mass index (BMI) of 25.9 ± 4.9 with a range of 17.7-32.6. Exclusion criteria included the use of assistive walking devices, obesity, pain in the leg or hip, history of lower limb trauma or surgery, or other physical conditions and pathologies that would affect gait or the ability to complete the 3-4 trials consisting of 3 strides each. One subject’s trials were excluded due to a malfunction with the data recorder. All protocols were IRB approved through Clemson University and the Greenville Hospital System, and all participants signed an informed consent form with a full description of the study.

Calculations

The angular velocities of the shank, ω_s, and thigh, ω_t, measured with the gyroscopes were processed to allow direct comparison with the parameters available with the camera-based system. The absolute angular orientation of the shank, θ_s, and thigh, θ_t, were calculated using Matlab ® 7.0 to integrate the angular velocity from each gyroscope. The flexion angle of the knee, θ_{flex}, was thus calculated:

$$\theta_{flex} = \int(\omega_t - \alpha) - \int(\omega_s - \beta) + \theta_{start} \quad (1)$$

when α = average ω_t at rest and β = average ω_s at rest. The knee flexion angle at rest (θ_{start}) was measured by a goniometer while the subject was standing with knees in full extension.

The accelerometer's sensitivity to gravity, $\vec{g}$, was accounted for by subtracting the projection of $\vec{g}$ on $A_{x'}$ and $A_{y'}$. $A_{z'}$ was disregarded due to the negligible effect of gravity along the medial/lateral axis. Corrected acceleration values were expressed as $A_{x''}$ and $A_{y''}$:

$$A_{y''} = A_{y'} - \overline{A}_{y'} * \cos\theta_s \quad (2)$$

$$A_{x''} = A_{x'} - \overline{A}_{y'} * \sin\theta_s \quad (3)$$

where $\overline{A}_{y'}$ is equal to the average $A_{y'}$ during one second with the subject at rest.

Data Comparison

The accuracy of the inertial sensors compared to the camera-based system was evaluated based on continuous measurements during individual gait cycles as well as specific events in the cycle. By comparing points based on particular gait events, any errors due to the synchronization of the two systems were minimized. These events include peaks in angular velocity of the shank and thigh during the swing phase, angular velocity of the thigh during hip flexion and extension, and knee flexion angle. The combination of these parameters will test values that are directly measured by the gyroscopes such as angular velocity as well as the knee flexion angle that has to be calculated from the inertial sensors' measurements.

There were two methods used to compare peak values between the two systems: one was calculation of the root mean squared error of all the trials at a specified point, and the second was the Altman and Bland method. The root mean squared error (RMSE) gives an average difference between values for the two systems for a specific gait event for all the trials. The Altman and Bland method plots the average value of the two systems' measurements against the difference between the two systems [13]. This method allows for the determination of bias between methods as well as any relationship between the differences between values obtained the two systems and the mean values.

The RMSE and Altman and Bland methods give information on the relationship between two systems at a specific point, but to further examine the claim of equivalence between the two systems a comparison throughout the entire cycle must be made. The Passing and Bablock technique involves using linear regression to determine the equivalence between two methods. For the parameters shank angular velocity, thigh angular velocity, knee flexion angle, shank angular acceleration, and linear acceleration, a plot of the camera-based system vs. inertial sensors allows for the best fit line of the expression $y = \beta x+\alpha$ for each trial. There are three hypothesis tests that need to be met to justify equivalence based on this model [14]:

(1) The relationship between the two methods must be linear with correlation value (R^2) being nearly equal to one;

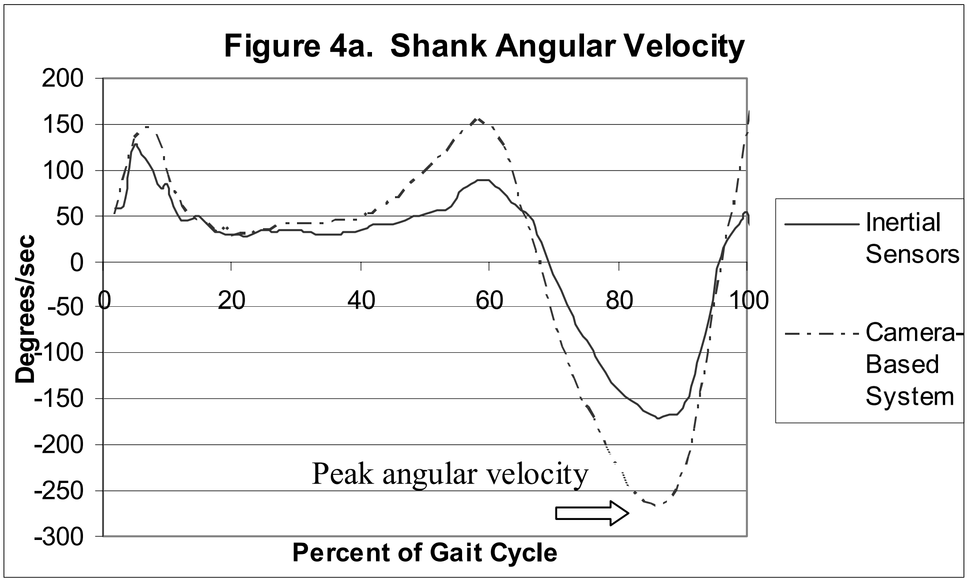
(2) The slope of the equation (β), must be equal to zero to eliminate the possibility of a proportional difference;

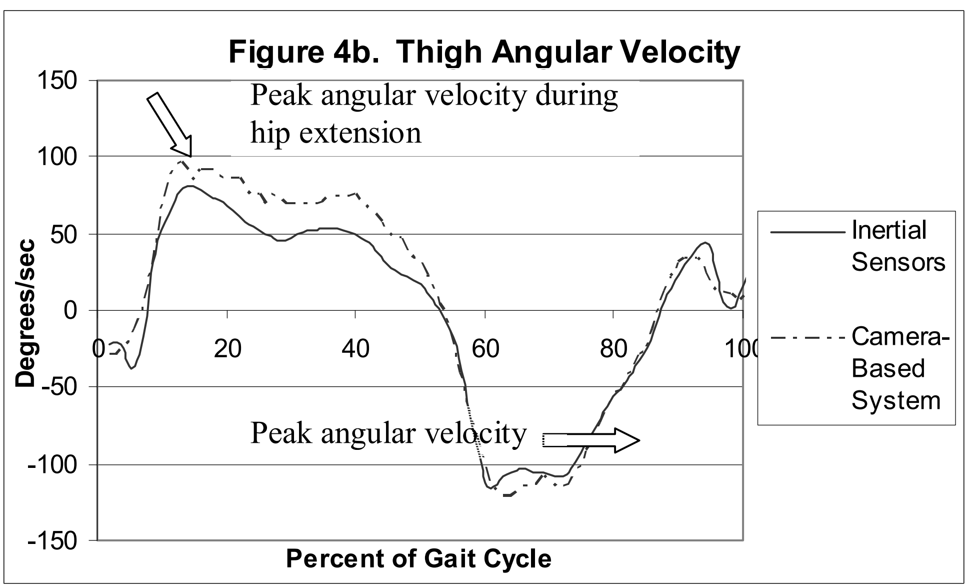
(3) The y-intercept (α), must be equal to zero to eliminate the possibility of a constant error.

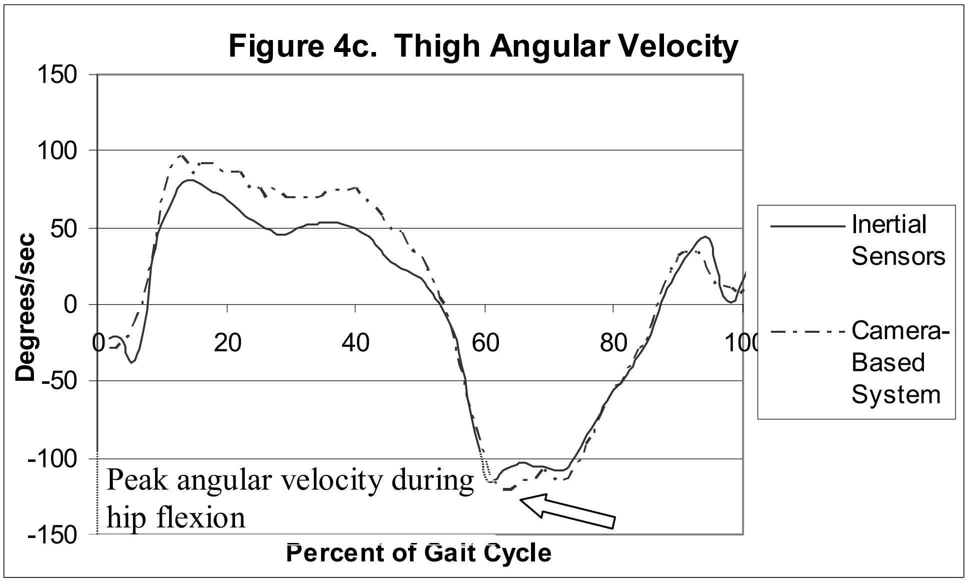
Results

The values of shank angular velocity, thigh angular velocity during hip extension and flexion, and knee flexion angle were compared at their peaks and throughout the gait cycle. (Figures 4a-4d).

Figure 4a.-4d. Peak Values on Superimposed Graphs of the Two Motion Analysis Methods







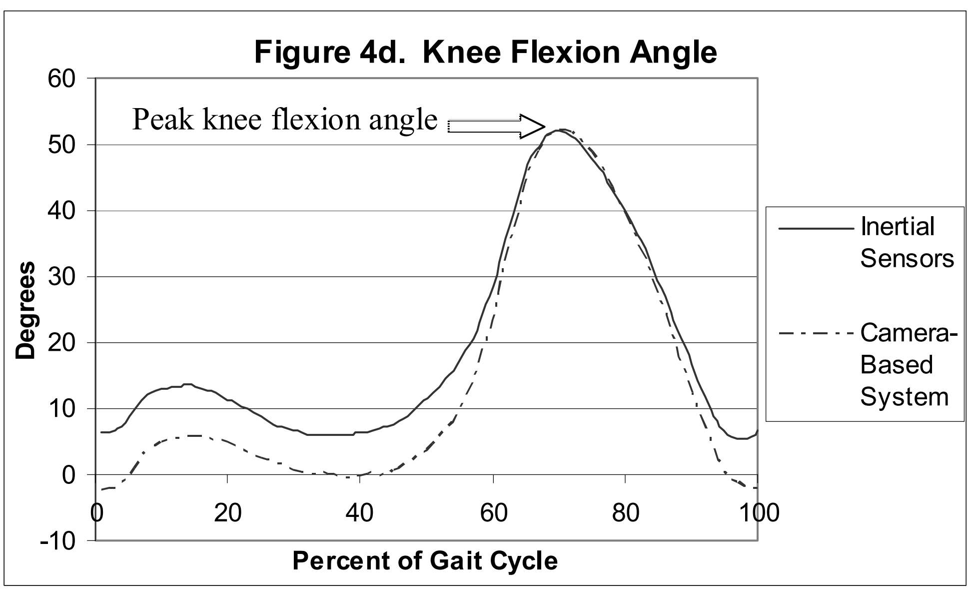


Figure 4a-4d. Superimposed graphs of the two systems. The arrows show the peak values measured by the gyroscopes and accelerometer.

Root mean squared error

The calculated root mean squared error (RMSE) for the peak angular velocities and knee flexion angle are shown in Table I. RMSE for the thigh during hip extension and flexion are less than the RMSE of the shank's peak angular velocity during the swing phase.

Table I. Root Mean Squared Error of Peak Angular Velocities and Peak Knee Flexion Angle

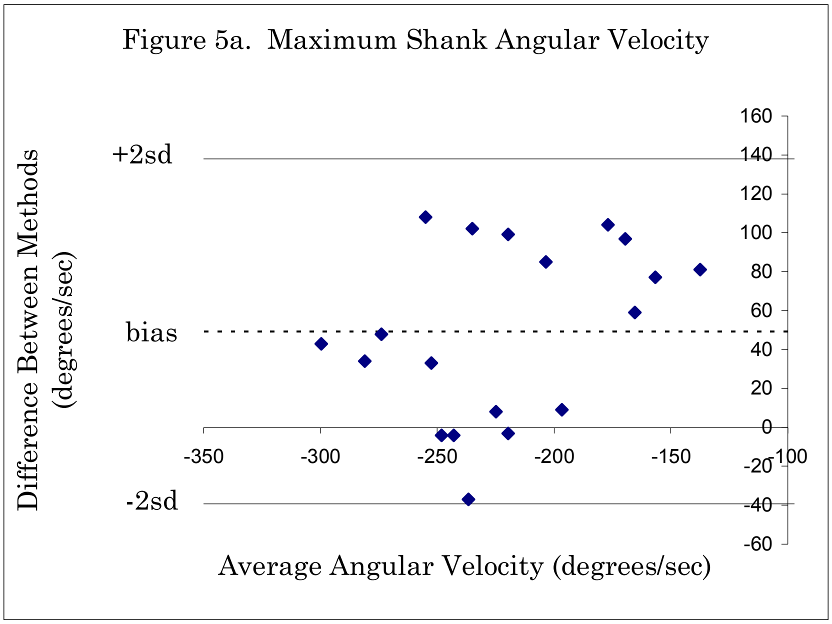
Variable	RMSE
Shank angular velocity	66.2 degrees/sec
Thigh angular velocity during hip extension	12.5 degrees/sec
Thigh angular velocity during hip flexion	13.2 degrees/sec
Knee flexion angle	7.9 degrees

The root mean squared error values used to compare the two systems at peak angular velocities and flexion angle.

Altman and Bland Comparison Method

The Altman and Bland method compared the two systems for individual points during the gait cycle. The peak shank angular velocity showed a bias of 49.4 degrees/sec with a large deviation with respect to the magnitude of the measurement (Figure 5a). There is no correlation between the magnitude of the angular velocity of the shank and the difference between methods, but the large bias and calculation of a range of errors shows that the angular velocities measured with inertial sensors are consistently lower than those determined with the camera-based system. The thigh angular velocity during hip flexion shows a negative bias (Figure 5c), and measured smaller magnitudes of the angular velocity by the inertial sensors since this is a negative peak with the defined reference system. The bias of the angular velocity of the thigh during hip extension is close to zero. With one outlier, it is seen in Figure 5b that the differences are equally distributed and less than 20 degrees/sec. The maximum flexion angle plot (Figure 5d) demonstrates smaller differences between methods for larger knee flexion angles. The bias as well as the majority of differences for maximum knee flexion angle were negative, which coincides with the smaller values found in the plots of the angular velocities of the shank and thigh from the gyroscopes since they are used in the calculation of knee flexion angle.

Figure 5a-5d. Altman and Bland Plots for Peak Values



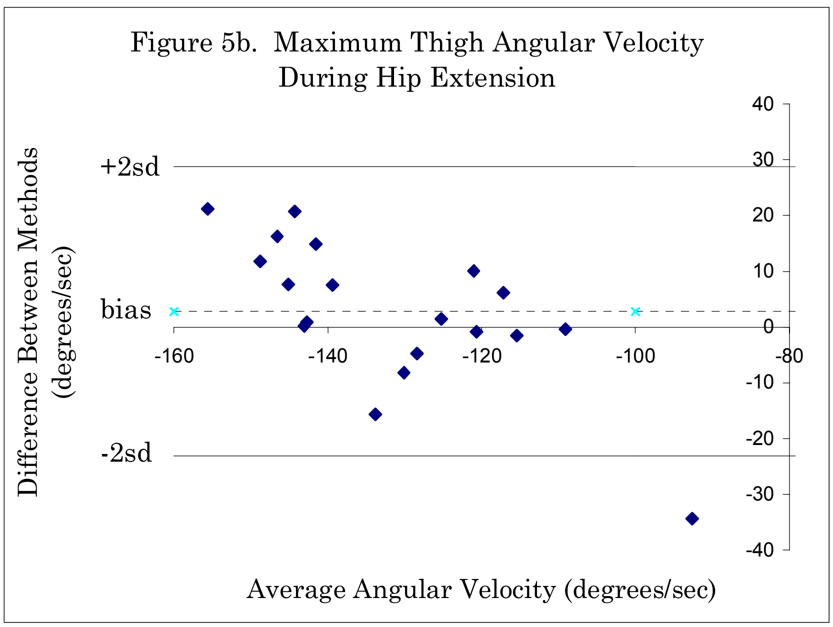


Figure 5c. Maximum Thigh Angular Velocity During Hip Flexion

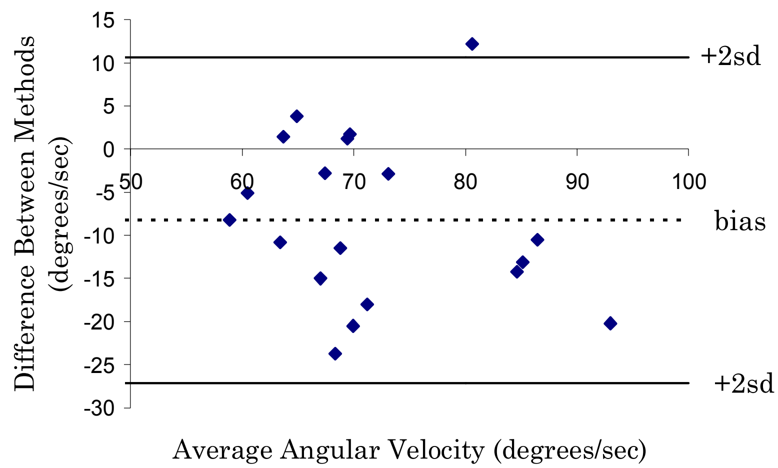


Figure 5d. Maximum Knee Flexion Angle

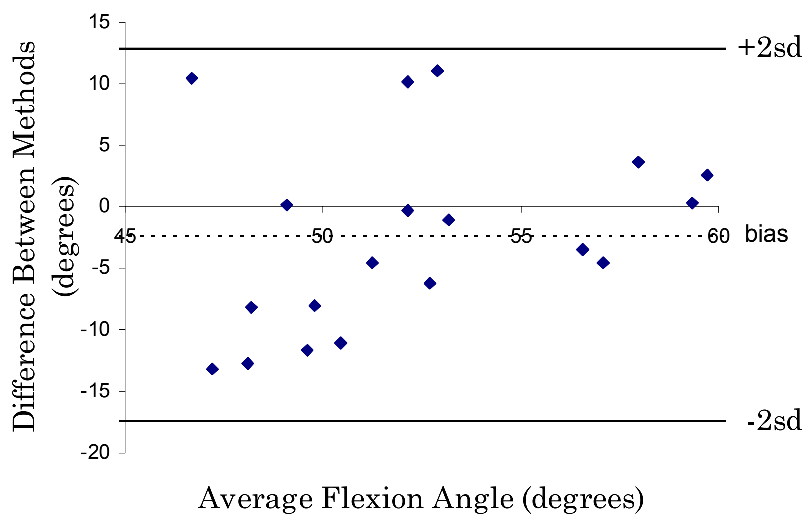


Figure 5a-5d. The differences between the values for angular velocities and knee flexion angle obtained with the two systems are plotted against the average of the two. The Altman and Bland method shows the range of 2 standard deviations from the average difference of the array of differences. Correlations of error with the magnitude of the measured value can be seen in these graphs.

Comparison of Gait Cycles

The Passing Bablok method was used to compare the two methods throughout a single gait cycle. Table II shows the average R^2, β, and α as well as their confidence intervals. It can be seen that the linear accelerations and angular acceleration of the shank have an R^2 that is much less than 1 and do not have a linear relationship. The slope and y-intercept for the angular velocities and knee flexion angle do not meet the equivalence requirements for $p \leq 0.05$.

Table II. Comparison of Gait Parameters by Linear Regression and the Passing Bablok Method

	Correlation Coefficient (R^2)		β (slope)		α (y-intercept)	
	Average ± s.d.	95% CI	Average ± s.d.	95% CI	Average ± s.d.	95% CI
Shank Angular Velocity	0.85±0.27	(0.73,0.97)	1.20±0.35	(1.04,1.35)	2.22±9.03	(-1.84,6.28)
Thigh Angular Velocity	0.93±0.08	(0.90,0.97)	1.05±0.07	(1.02,1.08)	5.20±2.42	(4.11,6.29)
Knee Flexion Angle	0.94±0.08	(0.90,0.97)	1.16±0.18	(1.08,1.24)	-6.63±5.35	(-9.10,-4.15)
Shank Angular Acceleration	0.41±0.27	(0.29,0.53)	0.65±0.37	(0.48,0.82)	26.8±86.3	(-12.0,65.5)
Linear Acceleration Ax	0.34±0.12	(0.27,0.40)	0.55±0.09	(0.50,0.59)	-2.26±2.08	(-3.35,-1.17)
Linear Acceleration Ay	0.19±0.12	(0.13,0.25)	0.22±0.20	(0.11,0.32)	-0.84±0.90	(-1.31,-0.36)

The slope and y-intercept of the best fit line when the inertial sensor data is plotted against the camera-based system data gives information on the differences between the systems.

The angular and linear accelerations of the shank do not meet the linear requirements for the Passing and Bablok method ($R^2 \neq 1$), but upon visual inspection of graphs with the two methods superimposed upon each other in Figures 6a-6c, it is apparent that the two systems show strong similarities for these variables in terms of overall shape and location of peak values in the gait cycle.

Figure 6a-6c. Superimposed Graphs of both Methods for Linear and Angular Accelerations

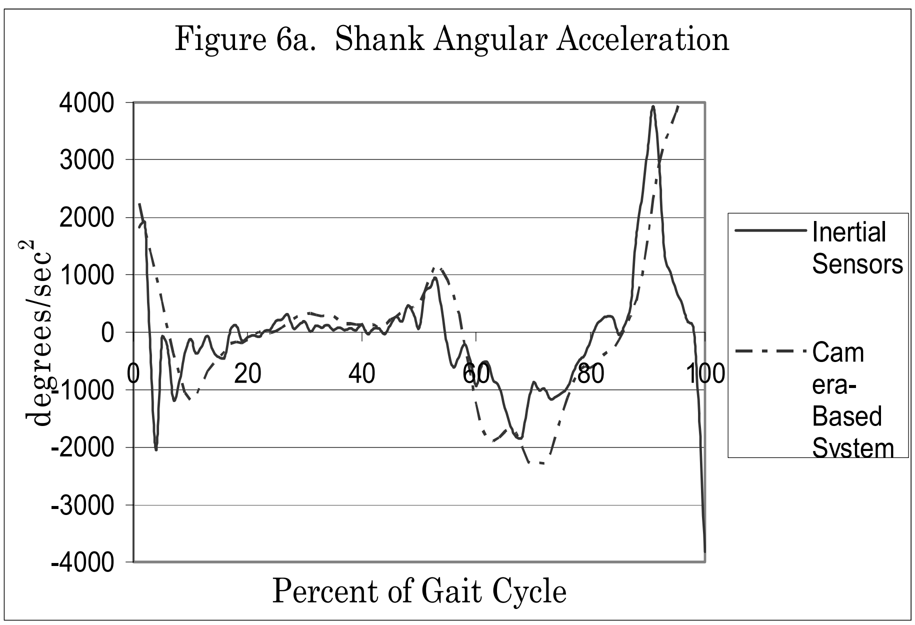


Figure 6b. Anterior/Posterior Shank Linear Acceleration

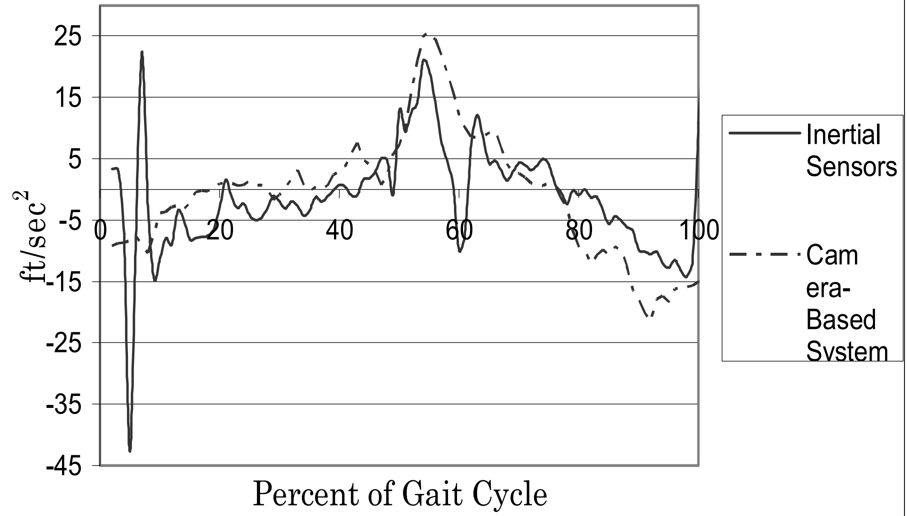
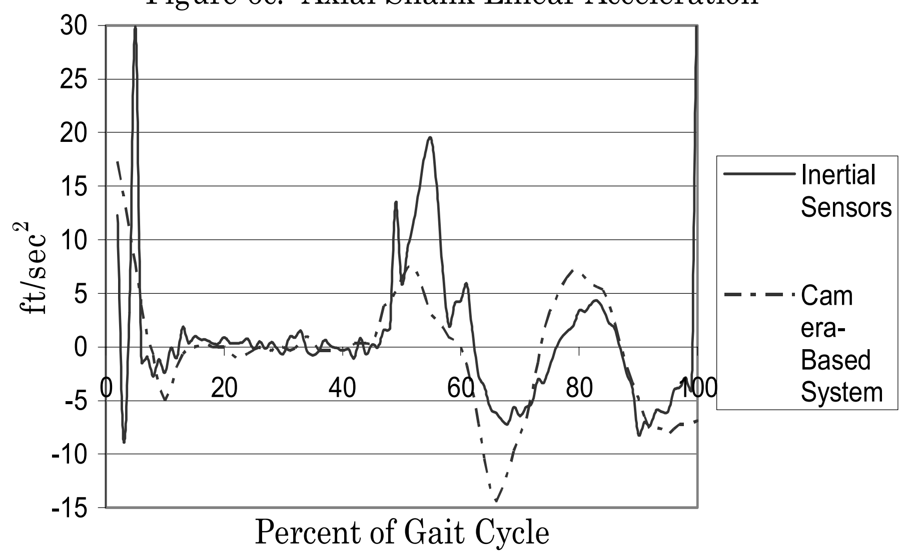


Figure 6c. Axial Shank Linear Acceleration



Discussion

The parameters selected for comparison in this study were done so because they test the "worst case" scenarios for both systems; namely, the direct measurement of position by the camera system compared to the calculated angular position with the inertial sensors, and the direct measurement of acceleration by the inertial sensors compared to the derived accelerations from the camera system data. One drawback of the inertial sensor system is that it cannot directly measure position; rather, linear and angular positions are obtained by integrating acceleration and velocity data, respectively. The conventional camera-based system directly measures position, and is thus likely to be more accurate than the inertial sensors. This is further amplified for knee flexion angle because of the cumulative effect of errors in angular position for the thigh and shank. In order to calculate the forces and moments on the knee, it is necessary to know the linear accelerations of the shank and the angular acceleration, respectively, so an error analysis of these parameters is necessary to account for possible weaknesses in the inertial sensor system.

A comparison of the two systems for angular velocities showed that the RMSE and the range of values in the Altman and Bland plot did not fall into an acceptable range for validation of the use of gyroscopes to measure angular velocity of the shank. The linear regression analysis gives some insight into the type of differences between the systems. There is a linear relationship between shank angular velocities for the methods, and the 95% confidence interval for the y-intercept includes 0, however, the 95% CI of the slope is (1.04 , 1.35) and does not include the necessary value of 1. This is consistent with results when there is a proportional relationship between the two methods.

The thigh angular velocities for the two methods demonstrated more similarities than the shank angular velocities. The difference between the methods for peak values in flexion and extension are approximately 13 degrees/sec, and have errors in a range that may still be useful for applications where an error of this magnitude would not effect interpretation. The linear regression statistics through the gait cycle show a positive y-intercept, which means a constant bias throughout the cycle for the thigh angular acceleration. Given these significant differences, it is not surprising that these are passed on to the knee flexion angle, which showed a proportional relationship and a bias to the results of the camera-based system because the measured values of angular velocity of the shank and thigh are used to calculate the knee flexion angle.

The angular and linear acceleration data from the inertial sensors had no filtering to smooth the lines. As a result, any comparison of the differences will show errors since the camera-based data is filtered [15]. The patterns of the inertial sensor data match that of the camera-based system; however there are some differences in values. During the toe-off phase [16], the accelerometers record high frequency peaks that could be missing from the camera-based data since it is filtered (Figure 6a and 6b). Since the inertial sensor system is designed for the evaluation of the kinetics of the knee joint, it is useful to have direct measurement of linear acceleration. Unfiltered data is beneficial to retain these peak values that are not captured with a camera-based system.

Errors

The two most common errors in inertial sensors when compared to a camera-based system is a lower value of angular velocity than that calculated from position data, and the noise associated with the angular and linear accelerations [17]. A third error that was reported is a time delay in the gyroscope data that creates a phase shift between the graphs of the two methods. This would account for some error in the angular velocities and knee flexion angles compared between the two methods at the designated time stamps, particularly in the linear regression analysis. This time delay has been reported in other studies and should only be a source of error when comparing to another system.

The smaller shank and thigh angular velocities have also been previously reported [9]. This error can be associated with the type of gyroscopes used in this study. Because uni-axial gyroscopes were used, no angular motion occurring outside of the sagittal plane is accounted for. While misalignment of the gyroscopes is a possibility, it is more likely that three-dimensional rotation of the shank results in lost data outside of the sagittal plane. Although a tri-axial gyroscope will add to the cost and size of the sensors, it appears necessary to be able to account for the rotation and any misalignment of the gyroscopes on the shank and thigh [5].

Previous studies have shown the ability of inertial sensors to consistently repeat accurate measurements when tested on motions with known values [18]. While the ability of the inertial sensors to report accurate measurements is not a source of error, it is possible that the sensors are moving during the trials. This would result in the sensors measuring accurate angular velocities and linear accelerations but not longer in the expected directions. The Coban wrap prevented checking the placement of the sensors between each trial, but the pressure of the sensors against the skin left an impression which was the same as the final placement of the sensors.

The noise of the linear and angular acceleration data limits the approaches of numerical comparison between the two methods without loosing information through filtering. While filtering can be useful to eliminate artifacts, there is a trade-off in loss of information. An alternative technique to filtering for improving the data quality includes using a smaller accelerometer to decrease the motion artifacts caused by the mass of the accelerometer and the distance it extends from the leg.

Conclusion

Inertial sensors are a viable alternative to camera-based systems for flexibility in the environments in which data can be collected. Strong similarities exist between the camera-based system and this particular arrangement of inertial sensors. From the comparison results, it can be concluded that with the addition of tri-axial gyroscopes, appropriate filtering of gyroscope data, and reduced size of the accelerometers, the errors can be minimized. There will naturally be some differences between the data collected by the two systems since they differ in which values are directly measured and which are found through calculations. Correction of the system's errors will provide a system that has the ability to measure or calculate the parameters needed to determine the forces and moments at the knee, while still having the versatility to be used in a clinical setting. Such a system could be used to track rehabilitation for individual patients and provide gait information for patients with a variety of lower limb pathologies and injuries without the need of a dedicated gait laboratory.

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CHAPTER 2

DISCUSSION

Inertial Sensors

Through the use of accelerometers it is possible to measure the linear accelerations of lower limb segments. Accelerometers function through piezoelectric materials and semiconductors in conjunction with a suspended mass in the material. When a force is applied to the mass it creates a change in the electric charge that results in a change in the voltage output [2]. Gyroscopes contain a small triangular prism that rotates around its axis in proportion to the angular velocity of the sensor [1]. More specifically, the Coriolis force acts upon the gyroscope to give the angular velocity. Recently there have been advancements in microelectromechanical systems (MEMS) and their application to inertial sensors. This technology integrates mechanical elements, sensors, actuators, and electronics onto a single chip, and dramatically decreases the size of the sensors.

Calculations

The gyroscopes and accelerometer give a direct measurement of angular velocity and linear accelerations. From these parameters, it is possible to use simple calculations to determine values typically determined from other motion analysis systems such as orientation of the thigh and shank, as well as their angular acceleration. In order to calculate angular acceleration, we use the equation:

$$\frac{d\omega_s}{dt} = \dot{\omega}_s \tag{5}$$

to calculate the derivative of the angular velocity from the gyroscope. Because the angular velocity of the shank is not filtered, the angular acceleration data has a substantial amount of noise. A ten point moving average was used to smooth the data in this study. The orientation of the shank and thigh is calculated by the following equation:

$$\omega_s dt + \theta_{s_{initial}} = \theta_s \tag{6}$$

and does not need to be smoothed since the calculation uses integration. The gyroscope is used to calculate the angle of a segment relative to its starting position. A goniometer was used to measure the initial angle of the shank before movement so that the angle of the shank relative to a global coordinate system can be given.

Comparison of Methods

There are many benefits to using an inertial sensor system for motion analysis. One advantage of accelerometers and gyroscopes is their small size and negligible mass in comparison to the limb of a patient. This will limit any influence that the sensor could have on the natural movements of the patient. Inertial sensors have the ability to be used as a long term monitoring system [3, 4] because the power supply is small enough to be run by a battery, and a portable data logger can be used to store data. Gyroscopes and accelerometers directly measure angular velocity and linear acceleration, respectively, which are important for understanding the kinetics of the knee. Additionally, the cost of an inertial sensor system is considerably less than that of a camera-based system. For gait analysis to become a tool that is regularly used by clinicians, it must be cost effective.

There are drawbacks to using an inertial sensor system for motion analysis. The gyroscopes directly measure angular velocity, which must be integrated in order to calculate angular orientation. Since integration requires a known initial condition, the subject must start motion with their lower limb in a known angular orientation. This can lead to errors in the calculation of angular orientation and flexion angle of the knee.

Current camera-based systems involve the use of multiple cameras and dedicated laboratory space. This makes these systems expensive and limits the scope of their use to a laboratory setting [1, 5]. For these types of tests, subjects are instrumented with reflective markers and walk across a force plate that is embedded in the floor. The range of motion for subjects is limited to the capture region of the camera, and therefore the motions performed do not occur in the natural environment where the tasks are performed. A benefit to using traditional camera-based systems is that they directly measure position, and do not require known initial conditions as with inertial sensors. Because distances and positions of body segments are tracked, it is straightforward for packaged software to visualize the movements of the body. The major drawback that results from directly measuring position is that inverse dynamics must be used to calculate accelerations from position data, which amplifies errors. Low-pass filters are typically used to remove sharp peaks that could be errors resulting from the differentiation steps. However such filters are unable to distinguish if peaks are due to motion or error, therefore, any peak values for velocity and acceleration are lost.

Errors Associated with Inertial Sensors

A common error associated with data from gyroscopes and accelerometers, particularly for integrated data, is drift over time and when the subject changes direction during motion [1, 6, 7]. There are several techniques that have been used to avoid this error. The test can be designed for a short duration, and with the subject moving in only one direction. An algorithm can be written to recognize heel strike and recalculate the tilt from the accelerometer at each heel strike to accommodate turning and a long duration test [8]. A third option is to monitor the starting and final positions and from this data account for any drift that occurs during the trial [9]. Finally, the data can be processed using filters to reduce the effect of drift [6].

A less common problem is temperature effects on the piezoelectric material or semiconductors that comprise the accelerometer [2]. When used in an indoor setting the temperature will not alter the readings, and as technology improves, accelerometers increase their tolerance to temperature fluctuations.

The placement of sensors is crucial regardless of the system being used. Gyroscopes have the advantage of only needing to be aligned in the proper plane to record the desired data. Gyroscopes that are co-planar will give identical measurements; therefore, the location along the length of a segment can be adjusted to adapt to anatomical variations [7, 8]. A disadvantage of the uni-axial gyroscopes is that they must be placed with their sensitive axis along the desired plane and any misalignment or rotation will cause a loss of data. The placement of the accelerometer on the anterior tibia limits but can't eliminate soft tissue error. This skin movement artifact is an error that is also seen in camera-based systems, particularly during high velocity motions. Also, it is important to place the accelerometer at the estimated center of mass of the shank for proper rigid body dynamics calculations because when calculating the forces acting on the knee, the accelerations needed are at the center of mass.

Error Analysis

The ability to directly measure angular velocities and accelerations with inertial sensors provides certain advantages. However, when using inertial sensors in the place of a traditional camera-based system, there are errors that must be taken into account during the system's design. It is important to examine the errors between trials of a single subject as well as the average difference over all trials.

Table III. Intra-Subject Variability between the Inertial Sensor System and the Camera-Based System

Parameter	**Variability**
Knee Flexion Maximum	2.1°
Peak Shank Angular Velocity	17.7°/sec
Peak Thigh Angular Velocity (hip flexion)	5.6°/sec
Peak Thigh Angular Velocity (hip extension)	9.6°/sec

The intra-subject variability is a measure of the average variance of the differences between the two methods for each subject's trials. It makes no statement on the magnitude of the difference but rather compares the consistency of the errors for each subject.

For each subject, the variance of the differences between the two systems' values for knee flexion maximum, peak shank angular velocity and peak thigh angular velocity in hip flexion and extension was calculated. The average intra-subject variability will reveal information about the comparison of errors for a single subject over multiple trials. The small intra-subject variability values in Table III make the claim that within multiple trials for each subject, the errors remained similar. This supports the idea that the error can be caused by the gyroscope being rotated out of the plane of its sensitive axis during characteristics that are specific for a subject's gait.

There have been three sources of error reported that were found in these data sets. First, at heel strike the lower limb is subjected to the largest forces. It is during this time that markers would be affected by soft tissue deformation and slippage resulting in a motion artifact. During the trials with the accelerometers, a large peak was seen in the linear acceleration in all three directions at heel strike. This correlates with previous trials where at large accelerations the peak is deformed and overestimates the actual linear acceleration because of vibrations and slipping of the accelerometer [5].

Second, a phenomenon that has been seen in the comparison of gyroscopes to other devices is a time lag in the data from the gyroscopes [10]. This was also seen during these trials, and while it has no effect on the numerical values recorded from the gyroscope it will cause the differences between the two systems to be larger due to the phase shift.

Third, an error that seems to be the cause of the underestimation of the angular velocities of both the shank and thigh is due to three-dimensional rotation of the limb [11]. The gyroscopes only measure the angular velocity along the sensitive axis of the gyroscope. Rotation of the leg causes the sensitive axis of the gyroscope to no longer be in the sagittal plane. Assuming similar gait between trials for a single subject, the within subject variance would be small as is shown in Table III.

Recommendations to Correct Errors with Inertial Sensor System

The aforementioned errors can be minimized by making a combination of three changes to the system. First, recent technology has decreased the size of inertial sensors. The mass and size of the accelerometer will have an impact of the artifacts found due to the sensor slipping and vibrating. By replacing the accelerometer with a smaller sensor, the error due to its mass can be eliminated. Second, it has been mentioned that the starting position of the shank is measured by a goniometers since the calculation of its angle is relative to its initial angle. A more accurate representation of the orientation of the shank can be found by using the accelerometer's reading of gravity to determine this position[12]. The third correction would be to replace the single axis gyroscopes with tri-axial gyroscopes. Adding the additional sensitive axes would allow for rotation of the leg while still collecting accurate data.

Applications of Sensor System

Osteoarthritis

Osteoarthritis (OA) is a common disease that appears in all races and with two-thirds of the cases involving females [13]. It is reported that in 2002, 42.7 million Americans, which accounts for 20.8% of the population [14], suffered from some form of arthritis with osteoarthritis being the most common, approximately 20 million people [15]. The significant impact of OA is the healthcare costs, and that it is one of the five leading causes of physical disability in non-institutionalized elderly men and women [16, 17], and it costs the economy approximately $60 billion annually [15].

The medical costs of patients with osteoarthritis are significant. Over 61% of adults with OA received treatment for their conditions [13]. In 2003, there were over 18.9 million visits to physicians by patients with OA in the knee, and Medicare paid for 47% of these visits [18, 19]. Of procedures performed on OA knee patients, the total knee replacement is the most common. In 1999, there were 249,000 total knee replacements in OA knee patients [16]. Osteoarthritis is a disease whose occurrence is proportional with age. As baby boomers continue to grow older, it is likely that an increase in prevalence is likely. It is projected that by the year 2020 there will be an estimated 57.5 million adults that will have doctor-diagnosed arthritis [20].

The debilitating effects of the pain associated with osteoarthritis leads to hidden costs to society such as missed work and the necessity of assistance to complete daily tasks. In 2003, more than 40% of adults with arthritis said that the pain interfered with their daily function activities [20]. This pain can often lead to missed days of work. Adults with OA missed over 13 Days of work due to health reasons, while adults without OA reported missing just 3 days.

The prevalence of osteoarthritis (OA) has led to studies examining the characteristics involved with lower limb kinematics in this group. A common symptom of osteoarthritis is some degree of varus/valgus misalignment that often leads to pain in the joint, whether it is the hip or knee. Numerous studies have examined the spatio-temporal, kinematic, and kinetic changes associated with knee OA [21-23]. The results are similar to other disorders that result in lower limb pain. Kaufman et al have shown that OA patients, when compared to a normal group, show a decrease in their average maximum flexion angle of 54.7° compared to 60.4° in the normal group [23]. Other studies have shown that the smaller maximum knee flexion angle found in OA subjects also had lower values for their knee flexion at heel strike and during the stance phase of the gait cycle [24]. In addition to the kinematic changes, there have been important changes noted in the kinetic parameters of OA patients. Using a multivariate gait data analysis technique, Astephen and Deluzio were able to use discrimination to determine the power of features in distinguishing normal and OA gait. Some of the most important factors in distinguishing OA gait includes: a lower medial-lateral contact force in the knee, a larger knee adduction moment and a knee internal rotation moment that that increases at a lower rate during the stance phase of the gait cycle [22]. There has also been evidence of a decreased strength in the knee extensors, a smaller knee extension moment and decreased vertical ground reaction force peak, but the loading rate to that peak is greater for OA gait [24]. It has been theorized that the lower knee extension moment is the result of a change in the muscle activation patterns during the gait in response to pain [24]. If this is the case than the decreased lower knee extension moment is important for the stabilization of the knee.

Because of the common clinical occurrence of OA, there are many scoring systems using measurements as well as pain evaluations to give a value to knee function. The inertial sensor system along with a force plate would have the ability to calculate the knee flexion angles of the knee, but also directly measures linear acceleration and calculates angular acceleration with a first order derivative which are both needed for the calculation of the forces and moments of the knee that distinguish the two gaits.

Lower Limb Pain

Lower limb fractures include fractures of the tibia, fibula, femur, foot and ankle. In 2003, there were approximately 91,000 reported fractures of the tibia/fibula, 422,000 femur fractures, and 170,000 fractures of the foot and ankle [25] . Many of these cases, particularly fractures of the femur and tibia result in surgery to correct the instability of the fracture.

Two common methods of fracture fixation are to use a plate or intramedullary nail to stabilize the fracture site and absorb some of the load, respectively. A part of these treatments involve early load bearing to facilitate bone formation and healing, but must be done in a manner that does not subject the patient to their full weight immediately after surgery.

When a patient experiences lower limb pain, including the hips and knees, there is a visible alteration in their gait. Because early weight bearing is often encouraged, the use of a motion analysis system for gait analysis provides a way to quantify the changes in kinematics and kinetics. Previous studies have examined the difference between groups with or without lower limb pain for parameters such as: walking velocity, range of motion of the knee and moments of the knee and hip [26]. It has been found that the compensatory measurements involved results in a slower gait with less flexion/extension of the knee. These mechanisms in return, lower the internal forces on the knee and hip to reduce the pain [26]. The same techniques have also been applied to subjects with and without patello-femoral pain (PFP) during activities of daily living (ADL) such as walking up and down stairs [27]. The importance of examining subjects during numerous types of experiments is shown by the variability of compensation that occurs between subjects with pain walking up stairs versus walking downstairs. The study showed a significant slowing of the cadence while descending stairs as well as a significant decrease in the initial peak of the knee extensor moment with the PFP groups having a peak of 0.75 Nm/kg and the normal group recorded an average value of 1.11 Nm/kg. The steps dictate the angle of the knee; therefore the decrease in the moment is predicted to be due to the muscle activation or an alteration of the center of mass.

Total Knee Arthroplasty

It has become common to examine the postoperative effects that a procedure has on the patient's kinetics and kinematics [28]. The complex mobility required from an implant in a total knee replacement makes this procedure a strong candidate for this type of testing. There are many knee-scoring systems that are used to evaluate function, but gait analysis provides quantitative results [21, 29]. Because the interest is on the improvements provided by the procedure, patients with a form of disability are tested in the pre and postoperative states to examine the effect of the procedure [28, 30].

In order to allow proper healing time, these studies are often conducted at different time intervals ranging from 4 months to 2 years postoperative [28, 31]. While several studies note the decrease in pain after the TKA, the operation appears to show little improvement in gait kinematics [28]. Studies have shown TKA patients 1-year post-op with significantly below normal knee range of motion and extension [30, 32]. The kinetics show similar changes to the OA compensatory mechanisms. It has been reported that the peak knee extensor torque and peak knee flexor torque were between 60-70% of the normal group in the group with TKA. The decrease in the extensor torque was larger resulting in a lower extensor/flexor ration of

1.575 than the 1.809 of the normal group [30]. Evaluations have been done comparing different implant systems such as: posterior-retaining vs. non posterior-retaining implants [32], and constrained vs. unconstrained [33], to compare post-operative kinematic improvements.

When the patients in these studies reached the point of needing the TKA procedure the degenerative disease had gotten to the point where walking was painful. The surgeries did help their pain, however, the importance of attempting to restore the gait to normal values should not be over looked. An inertial sensor system can track the progress of a rehabilitation program in the areas that were found to still be different from normal 1-year post-op such as the knee extensor and flexor torques and the ROM of the affected knee. This would apply equally as well to compare the effectiveness of the joint replacement by these kinematic and kinetic properties.

Knee Scoring Systems

For assessment before and after treatments of a variety of conditions, scoring systems have become a popular tool for a multitude of knee disorders [34]. Systems such as the Lysholm scale, Cincinnati knee-rating system and Activities of Daily Living scale of the Knee Outcome Survey often taken into account functional limitations caused by the condition [35, 36], and also allows for pain to count for as much as 24% of the total score.

Functional limitations can be described in an objective manner, however, having a patient grade their level of pain adds a level of subjectivity to the score dependent on their pain threshold [37]. It has been shown that there is a high correlation between scoring systems that use kinematics to those that include subjective scoring, particularly in the Hospital for Special Surgery (HSS) Knee Rating Form [38].

Pain is important to consider in these scoring systems since it limits function and defines the disabling nature of the disease, however, it is still advantageous to have an objective, quantitative scoring system that uses the kinematics and particularly the kinetics from gait analysis trials. By reviewing changes in the ROM of the knee and the forces and moments acting on the knee the progression of a degenerative disease can be defined, the improvements of a rehabilitation program quantified and the results of a particular surgical option compared to the alternatives.

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CHAPTER 3

CONCLUSIONS AND RECOMMENDATIONS

The use of gait analysis as a diagnostic tool and as an objective measure of knee function led to an investigation of an alternative technique to perform this analysis. The traditional method of using reflective markers and multiple cameras to track the movement of body segments presents limitations because of the laboratory needed for this equipment. Additionally, changes in kinetics for patients with osteoarthritis, total knee replacement or another type of lower limb pain, when compared to a normal group, have been documented as a characteristic of the disorders. The camera-based systems are accurate in their measure of position, but require inverse dynamics to calculate the velocities and accelerations needed to determine the forces and moments of the knee, and can lead to errors.

The inertial sensor system of two uni-axial gyroscopes and a tri-axial accelerometer used in this study addressed the deficiencies of the camera-based system by providing a small, light-weight system that has the ability to be a portable system, and directly measures the angular velocities and accelerations of the thigh and shank. Validating the inertial sensor method against the camera-based method by simultaneously collecting data from the two systems addressed the strengths and weaknesses of the inertial sensor method. Data was compared between both systems for parameters that the camera-based method measures accurately such as global orientation and joint angles. The tendency of gyroscopes to have an integration drift error occurs when calculating orientation, and without modifying the integration by a process similar to adjusting for the drift with boundary conditions error would be expected. Angular velocities and accelerations that are measured by the inertial sensors but calculated by the camera-based method were also compared. The inertial sensors were able to measure linear acceleration of the shank without losing the high frequency accelerations that are filtered out of the camera-based system's data to allow for differentiation.

Visual inspection of the data superimposed on single graphs showed similar results for both methods for knee flexion angle, angular velocities of the shank and thigh along the sagittal plane, angular acceleration of the shank and linear acceleration of the shank. The unfiltered shank linear acceleration data showed some measured high frequency peaks that were filtered from the camera-based data's calculations. Root mean squared errors and Altman and Bland plots were used to compare the differences of important peaks of shank angular velocity, thigh angular velocity and knee flexion angles. The Altman and Bland plots had a large distribution of differences for shank angular velocity, however, the bias of the peak angular velocity of the hip in flexion and extension were both less than 10 deg/sec. The RMSE for the peak angular velocity of the thigh in flexion and extension were 13.2 and 12.5 deg/sec respectively. The Passing and Bablok method compared the entire waveform for all the collected parameters. While examining the peak knee flexion angle showed differences between the systems ranging from -13° to 10°, the Passing and Bablock method had an R^2 average of .94 and a slope of 1.16 which shows strong similarities over the course of the gait cycle.

From the examination of the results, some suggestions for alterations to the system were made. It is known that there is rotation at the hip and knee during the gait cycle; however, single axis gyroscopes were used for this system. It is recommended that tri-axial gyroscopes replace the single axis gyroscopes to account for the gait characteristics that caused a loss of data from the out of plane motion and subsequent underestimation of angular velocities. A second recommended change was to reduce the size of the tri-axial accelerometer attached to the shank. The advantages of directly measuring linear acceleration and not filtering the data is lost if motion artifacts are present in the data. Minimizing the size of the accelerometer will decrease the noise caused by the mass of the accelerometer at heel strike.

The inertial sensor system used in this validation study showed many similarities to the data from the camera-based system. Some differences were expected because both systems have their own error when using measurements and calculations to describe the subject's gait. These similarities open the possibility for a variety of future studies involving the sensors. Changing the types of gyroscopes and accelerometer used should further reduce the system's errors, but a future comparison of the modified system against a proven method would quantify the improvements in accuracy. Besides using multi-axis gyroscopes and smaller accelerometers further modifications to the system can improve its functional ability. The wires that connect the sensors to the data recording system tether the patient to a location. A system where a small data recording device could be carried on the patient that will either save the data or transmit it to a computer would allow for increased versatility for applications of the sensors.

The repeatability of the sensor placement by an examiner should also be tested to eliminate sensor placement as a source of error or to describe the error caused by it. Traditional camera based systems use palpation to estimate the location of bony landmarks to place the reflective markers on, but gyroscopes can be placed a different lengths along a rigid body but the plane of motion is estimated. The usefulness of the sensors are negated if the sensors cannot be removed and then reattached to the same subject with angular velocities remaining in the range of normal intrasubject variability between trials.

The comparison of these two systems provides an alternative method of evaluating a subject's gait. This type of a system could be useful for its applications in a clinical environment where objective characterizations of knee function can be made to track the progression or severity of disabilities, effectiveness of treatments and the improvements made during a rehabilitation program.

APPENDICES

Appendix 1

Sensor Study Data

Maximum Knee Flexion Angle

Trial	Inertial Sensors	Camera-Based System	Difference
Subject 1 trial 1	59.5°	59.2°	0.3°
Subject 1 trial 2	44.1°	52.3°	-8.2°
Subject 1 trial 3	49.6°	55.8°	-6.2°
Subject 1 trial 4	49.0°	53.5°	-4.5°
Subject 2 trial 1	61.0°	58.4°	2.6°
Subject 2 trial 2	54.8°	59.4°	-4.6°
Subject 2 trial 3	54.8°	58.3°	-3.5°
Subject 2 trial 4	59.8°	56.2°	3.6°
Subject 3 trial 1	51.9°	41.5°	10.4°
Subject 3 trial 2	57.2°	47.1°	10.1°
Subject 3 trial 3	58.4°	47.4°	11.0°
Subject 4 trial 1	52.0°	52.3°	-0.3°
Subject 4 trial 2	49.2°	49.0°	0.2°
Subject 4 trial 3	52.7°	53.7°	-1.0°
Subject 5 trial 1	45.8°	53.8°	-8.0°
Subject 6 trial 1	40.6°	53.8°	-13.2°
Subject 6 trial 2	44.9°	56.0°	-11.1°
Subject 6 trial 3	43.8°	55.4°	-11.6°
Subject 6 trial 4	41.7°	54.5°	-12.8°
Average ± S.D.	51.1 ± 6.4°	53.6 ± 4.6°	-2.5 ± 7.7°

Peak Shank Angular Velocity

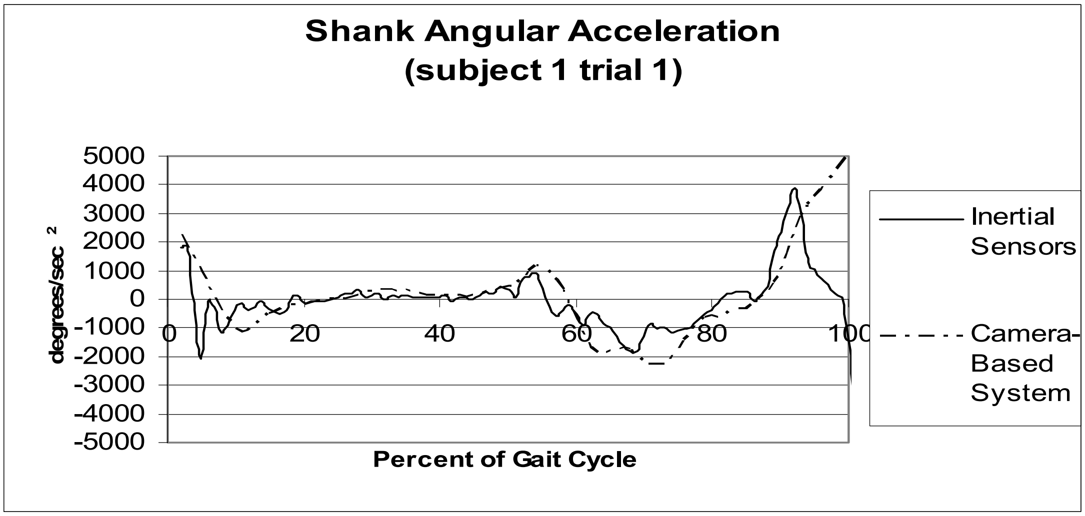
Trial	Inertial Sensors	Camera-Based System	Difference
Subject 1 trial 1	-184°/sec	-286°/sec	102°/sec
Subject 1 trial 2	-170°/sec	-269°/sec	99°/sec
Subject 1 trial 3	-201°/sec	-309°/sec	108°/sec
Subject 1 trial 4	-161°/sec	-246°/sec	85°/sec
Subject 2 trial 1	-236°/sec	-269°/sec	33°/sec
Subject 2 trial 2	-278°/sec	-321°/sec	43°/sec
Subject 2 trial 3	-264°/sec	-298°/sec	34°/sec
Subject 2 trial 4	-250°/sec	-298°/sec	48°/sec
Subject 3 trial 1	-221°/sec	-218°/sec	-3°/sec
Subject 3 trial 2	-245°/sec	-241°/sec	-4°/sec
Subject 3 trial 3	-250°/sec	-246°/sec	-4°/sec
Subject 4 trial 1	-192°/sec	-201°/sec	9°/sec
Subject 4 trial 2	-125°/sec	-229°/sec	104°/sec
Subject 4 trial 3	-221°/sec	-229°/sec	8°/sec
Subject 5 trial 1	-255°/sec	-218°/sec	-37°/sec
Subject 6 trial 1	-121°/sec	-218°/sec	97°/sec
Subject 6 trial 2	-136°/sec	-195°/sec	59°/sec
Subject 6 trial 3	-118°/sec	-195°/sec	77°/sec
Subject 6 trial 4	-97°/sec	-178°/sec	81°/sec
Average ± S.D.	-196 ± 57°/sec	-246 ± 43°/sec	49 ± 45°/sec

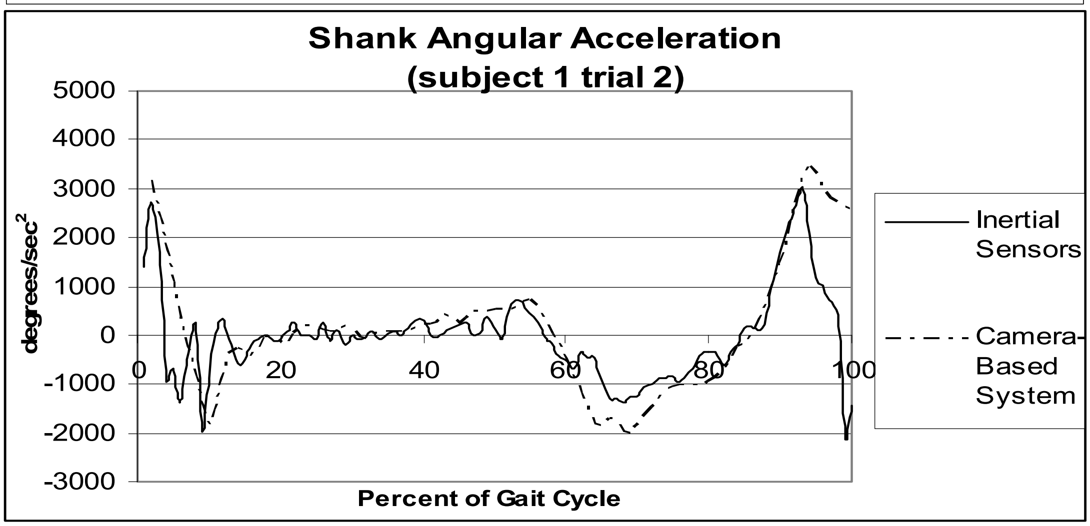
Peak Hip Angular Velocity for Hip Extension

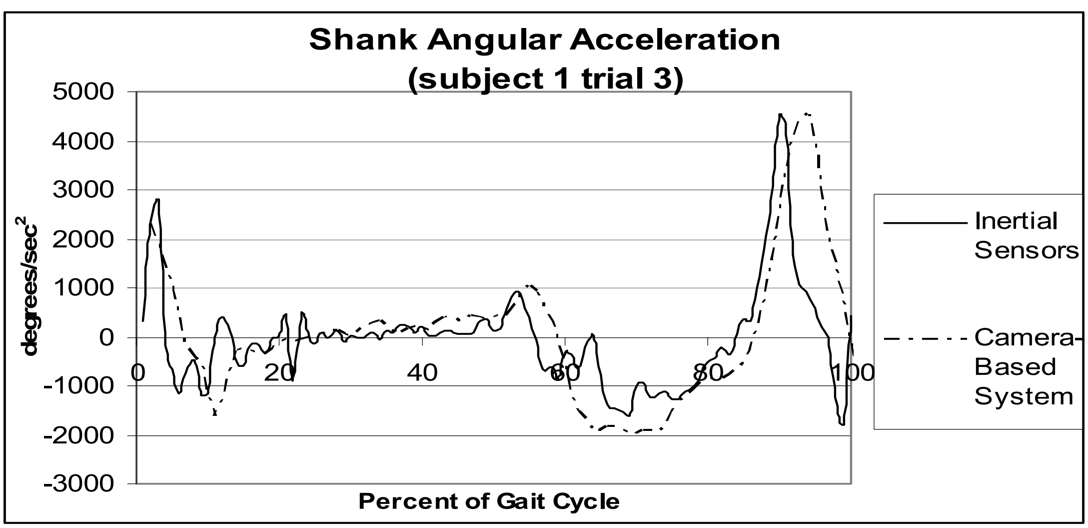
Trial	Inertial Sensors	Camera-Based System	Difference
Subject 1 trial 1	82.9°/sec	103.1°/sec	-20.2°/sec
Subject 1 trial 2	81.2°/sec	91.7°/sec	-10.5°/sec
Subject 1 trial 3	78.6°/sec	91.7°/sec	-13.1°/sec
Subject 1 trial 4	77.5°/sec	91.7°/sec	-14.2°/sec
Subject 2 trial 1	63.0°/sec	74.5°/sec	-11.5°/sec
Subject 2 trial 2	56.5°/sec	80.2°/sec	-23.7°/sec
Subject 2 trial 3	59.7°/sec	80.2°/sec	-20.5°/sec
Subject 2 trial 4	62.2°/sec	80.2°/sec	-18.0°/sec
Subject 3 trial 1	54.8°/sec	63.0°/sec	-8.2°/sec
Subject 3 trial 2	58.0°/sec	68.8°/sec	-10.8°/sec
Subject 3 trial 3	57.9°/sec	63.0°/sec	-5.1°/sec
Subject 4 trial 1	66.8°/sec	63.0°/sec	3.8°/sec
Subject 4 trial 2	70.0°/sec	68.8°/sec	1.2°/sec
Subject 4 trial 3	64.4°/sec	63.0°/sec	1.4°/sec
Subject 5 trial 1	59.5°/sec	74.5°/sec	-15.0°/sec
Subject 6 trial 1	86.7°/sec	74.5°/sec	12.2°/sec
Subject 6 trial 2	70.5°/sec	68.8°/sec	1.7°/sec
Subject 6 trial 3	71.6°/sec	74.5°/sec	-2.9°/sec
Subject 6 trial 4	66.0°/sec	68.8°/sec	-2.8°/sec
Average ± S.D.	67.8 ± 9.7°/sec	76.0 ± 11.6°/sec	-8.2 ± 9.6°/sec

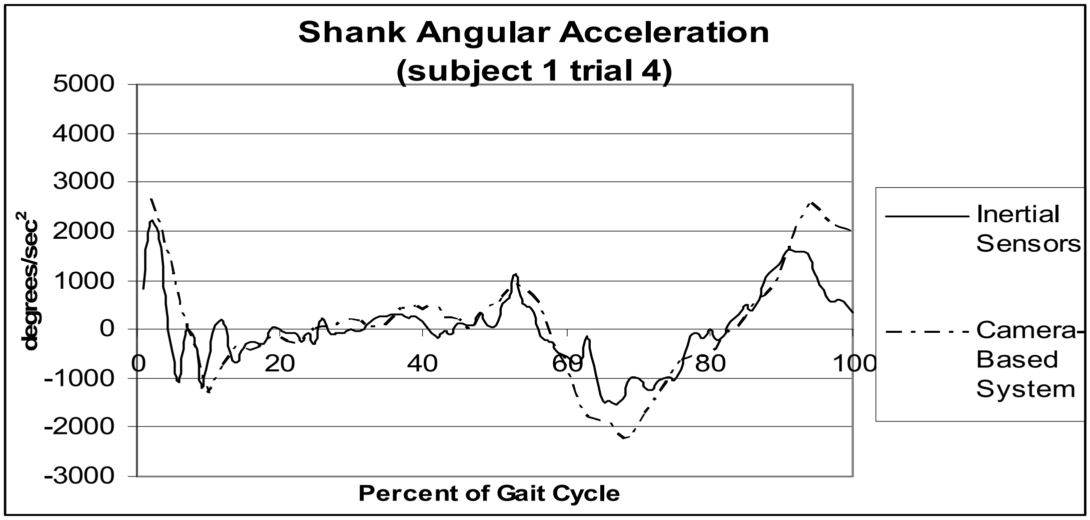
Peak Hip Angular Velocity for Hip Flexion

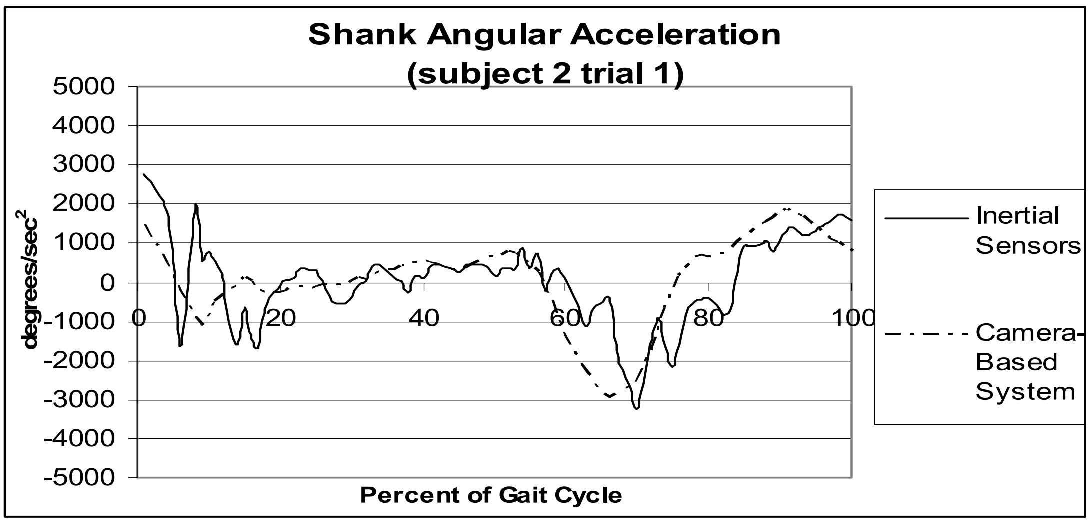
Trial	Inertial Sensors	Camera-Based System	Difference
Subject 1 trial 1	-145.0°/sec	-166.2°/sec	21.2°/sec
Subject 1 trial 2	-116.0°/sec	-126.1°/sec	10.1°/sec
Subject 1 trial 3	-133.9°/sec	-154.7°/sec	20.8°/sec
Subject 1 trial 4	-135.6°/sec	-143.2°/sec	7.6°/sec
Subject 2 trial 1	-141.6°/sec	-126.0°/sec	-15.6°/sec
Subject 2 trial 2	-141.3°/sec	-149.0°/sec	7.7°/sec
Subject 2 trial 3	-142.9°/sec	-154.7°/sec	11.8°/sec
Subject 2 trial 4	-142.3°/sec	-143.2°/sec	0.9°/sec
Subject 3 trial 1	-116.2°/sec	-114.6°/sec	-1.6°/sec
Subject 3 trial 2	-121.1°/sec	-120.3°/sec	-0.8°/sec
Subject 3 trial 3	-114.1°/sec	-120.3°/sec	6.2°/sec
Subject 4 trial 1	-134.1°/sec	-149.0°/sec	14.9°/sec
Subject 4 trial 2	-124.5°/sec	-126.0°/sec	1.5°/sec
Subject 4 trial 3	-138.5°/sec	-154.7°/sec	16.2°/sec
Subject 5 trial 1	-134.2°/sec	-126.0°/sec	-8.2°/sec
Subject 6 trial 1	-130.8°/sec	-126.0°/sec	-4.8°/sec
Subject 6 trial 2	-143.0°/sec	-143.2°/sec	0.2°/sec
Subject 6 trial 3	-109.3°/sec	-108.9°/sec	-0.4°/sec
Subject 6 trial 4	-109.8°/sec	-75.4°/sec	-34.4°/sec
Average ± S.D.	-130.2 ± 12.3°/sec	-133.0 ± 21.3°/sec	2.8 ± 13.2°/sec

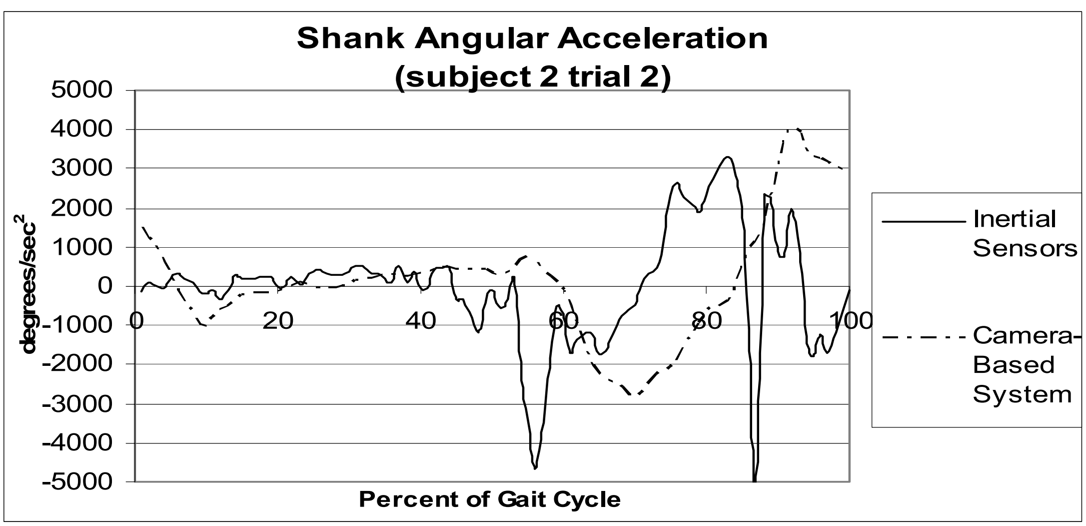

Shank Angular Acceleration
(subject 1 trial 1)
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
-4000
-5000
degrees/sec 2
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

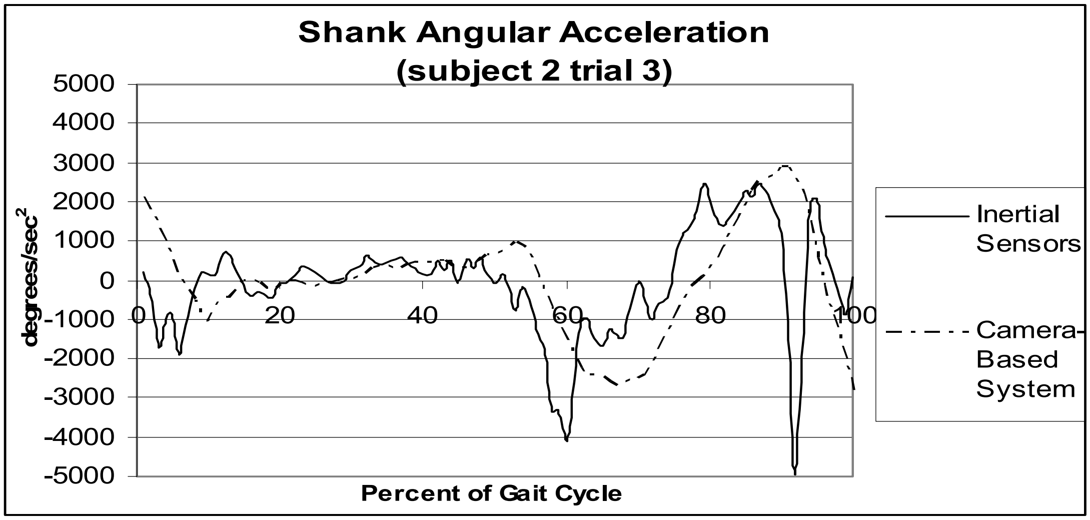

Shank Angular Acceleration
(subject 1 trial 2)
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
degrees/sec2
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

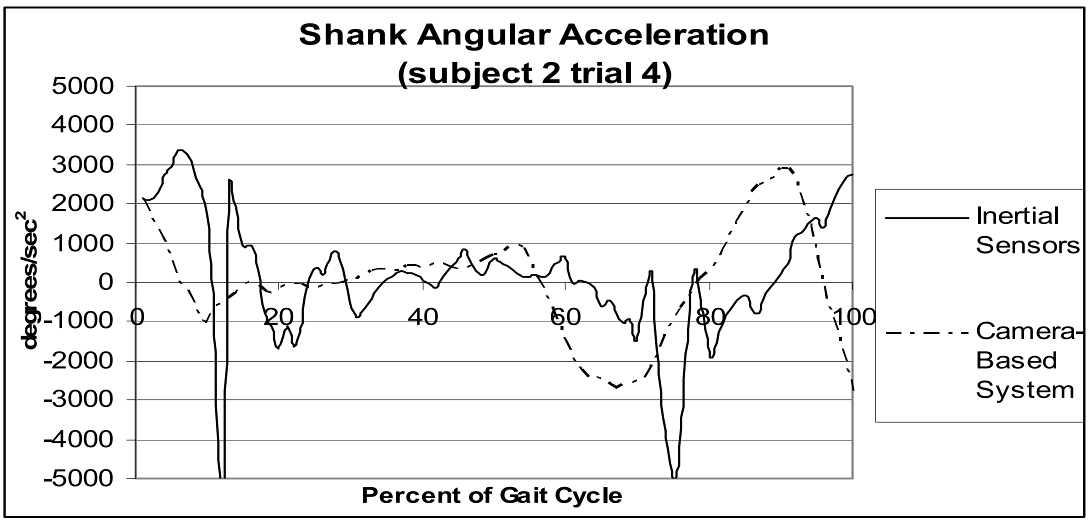

Shank Angular Acceleration
(subject 1 trial 3)
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
degrees/sec2
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

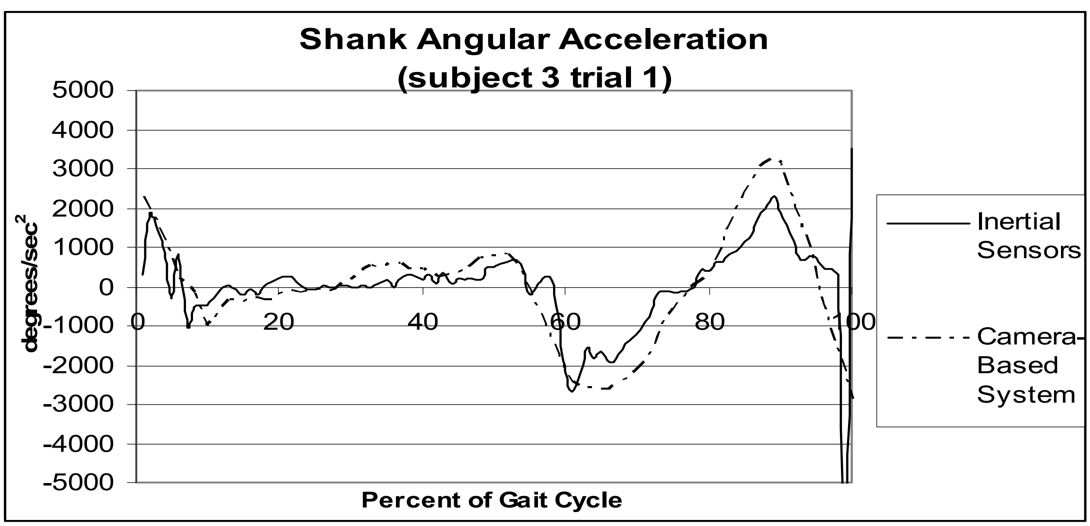

Shank Angular Acceleration
(subject 1 trial 4)
degrees/sec²
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

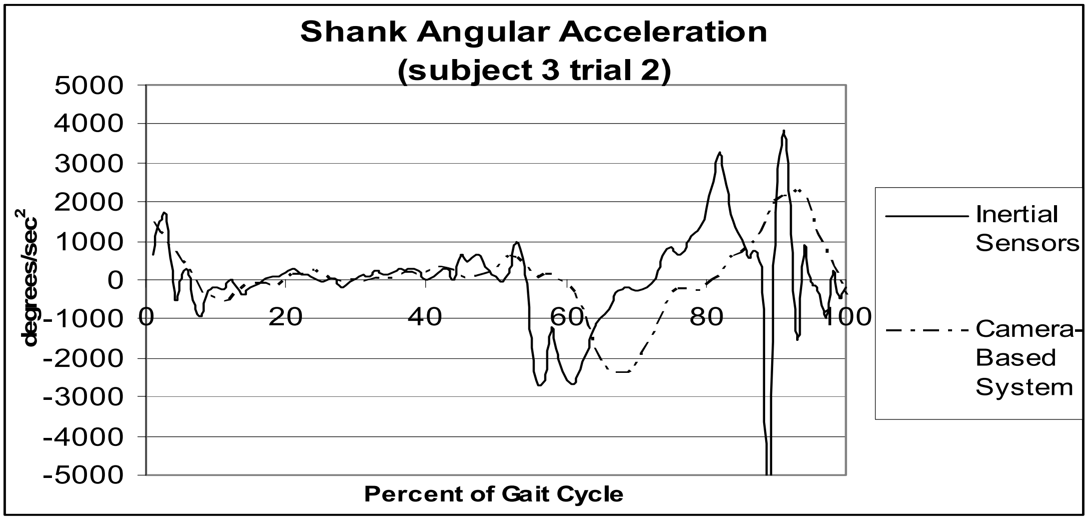

Shank Angular Acceleration
(subject 2 trial 1)
degrees/sec²
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
-4000
-5000
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

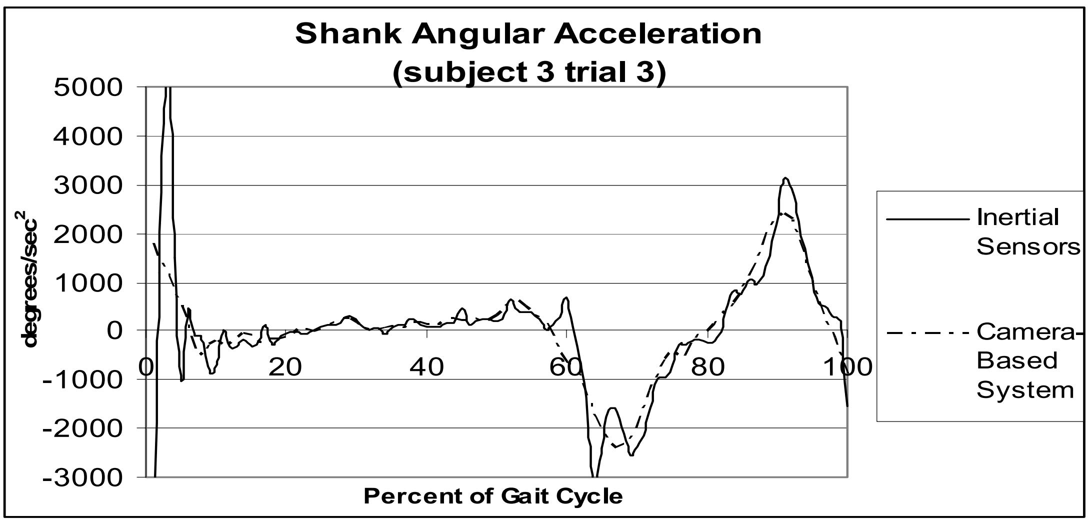

Shank Angular Acceleration
(subject 2 trial 2)
degrees/sec²
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
-4000
-5000
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

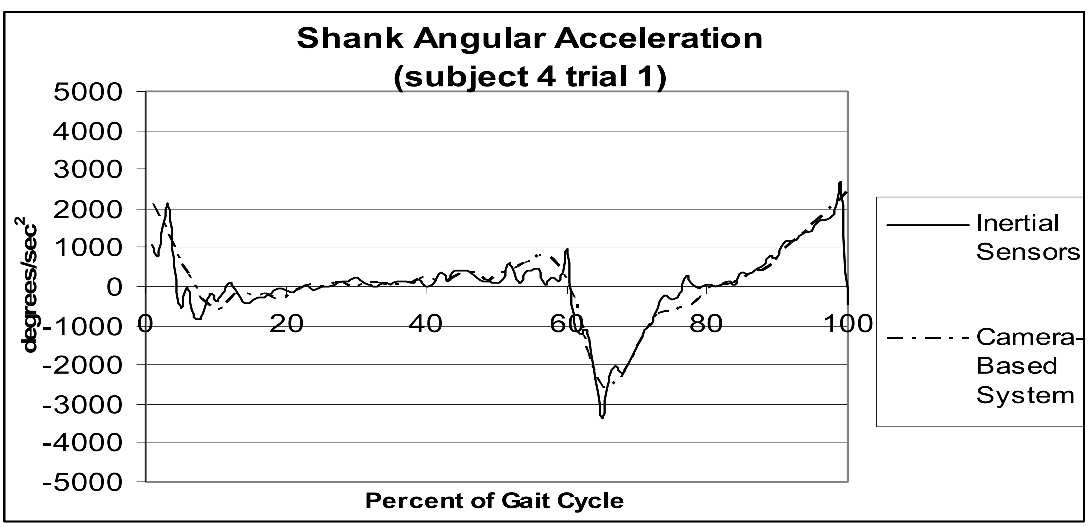

Shank Angular Acceleration
(subject 2 trial 3)
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
-4000
-5000
degrees/sec^2
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

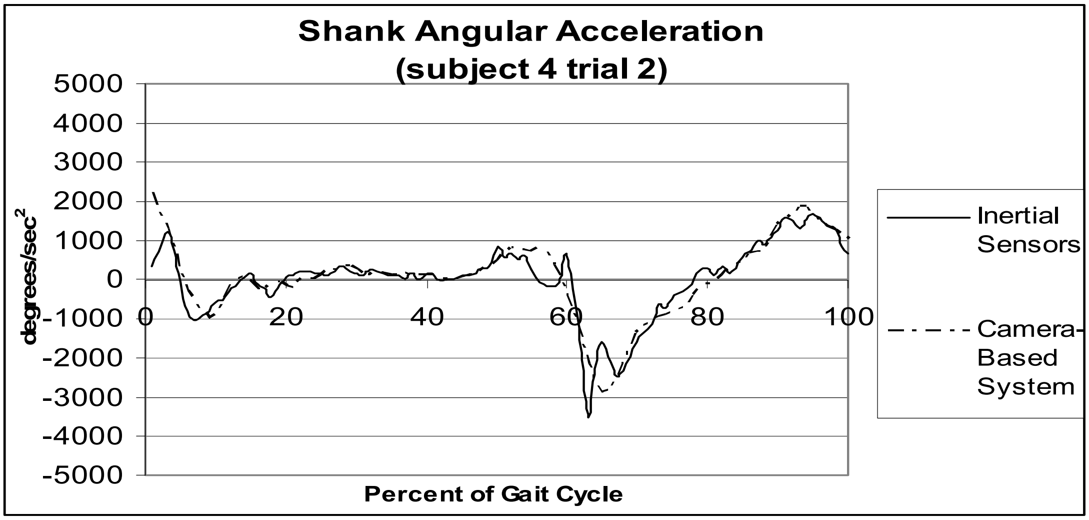

Shank Angular Acceleration
(subject 2 trial 4)
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
-4000
-5000
degrees/sec^2
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

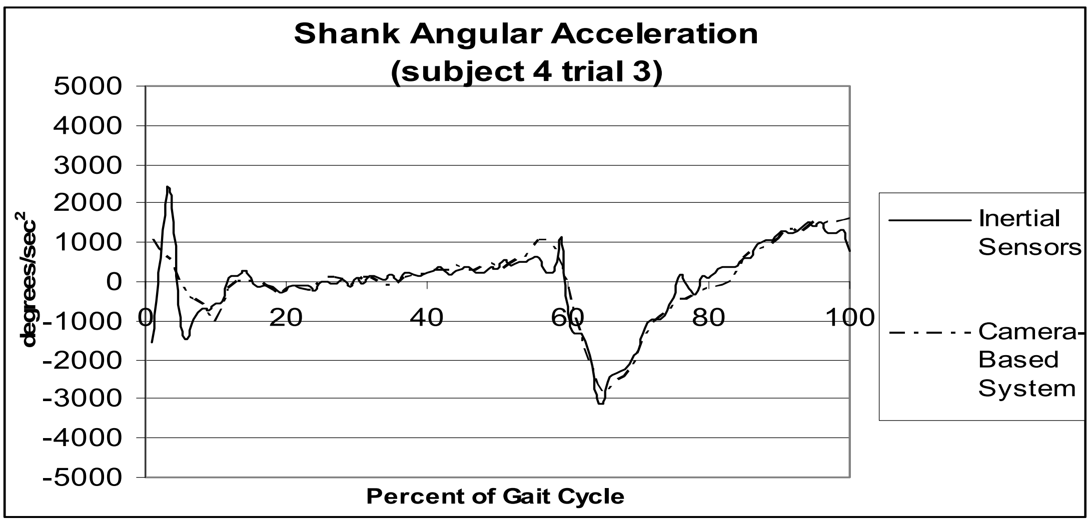

Shank Angular Acceleration
(subject 3 trial 1)
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
-4000
-5000
degrees/sec^2
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

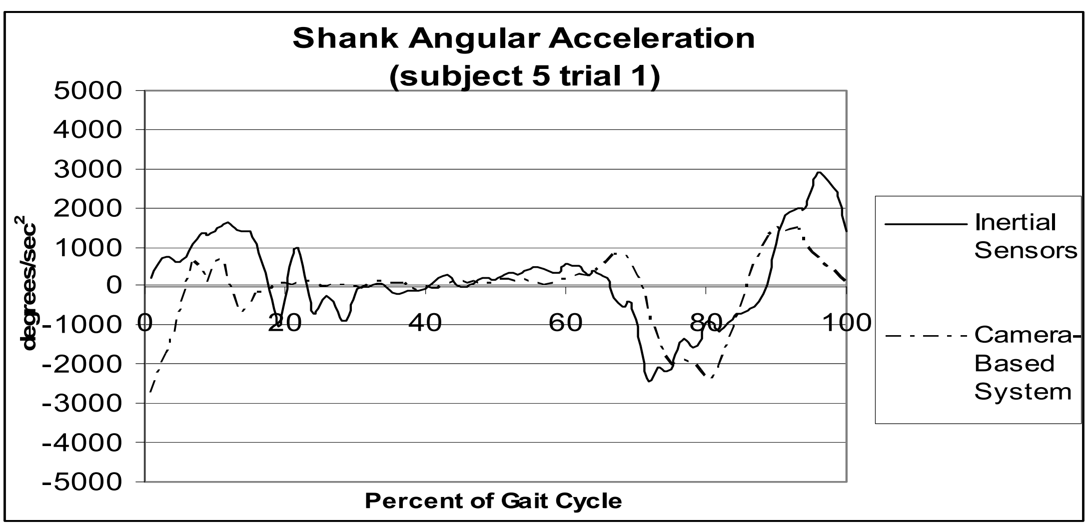

Shank Angular Acceleration
(subject 3 trial 2)
degrees/sec^2
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
-4000
-5000
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

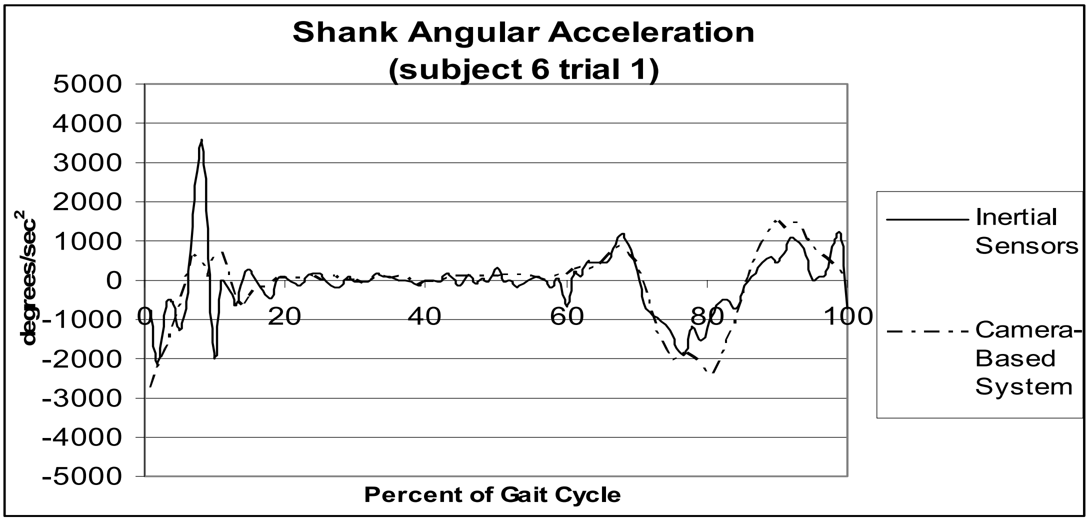

Shank Angular Acceleration
(subject 3 trial 3)
degrees/sec^2
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

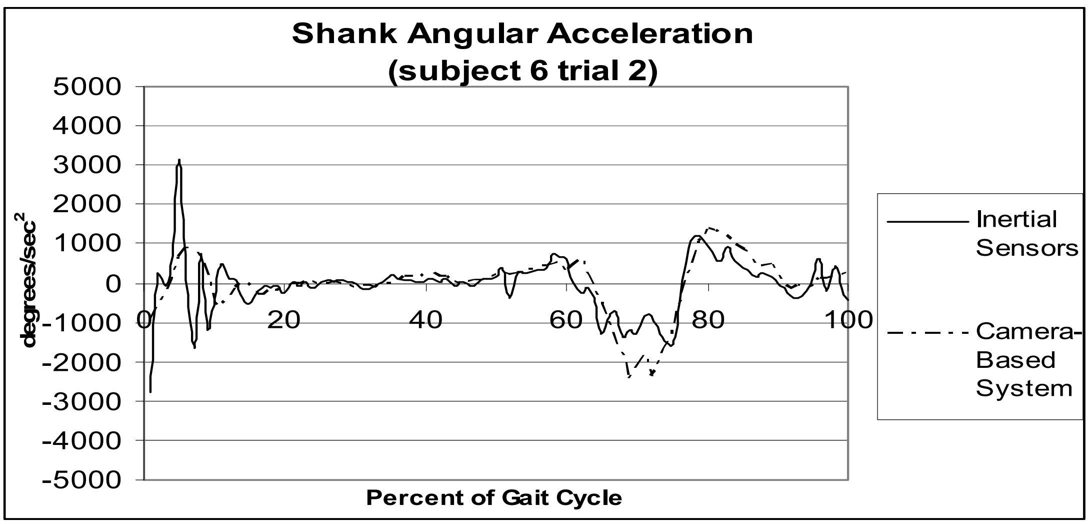

Shank Angular Acceleration
(subject 4 trial 1)
degrees/sec^2
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
-4000
-5000
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

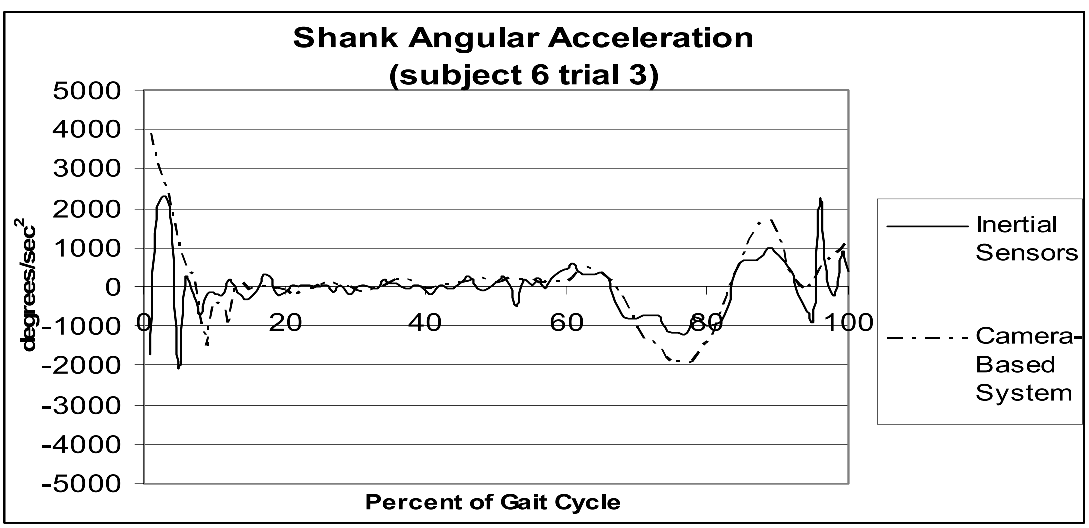

Shank Angular Acceleration
(subject 4 trial 2)
degrees/sec^2
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
-4000
-5000
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

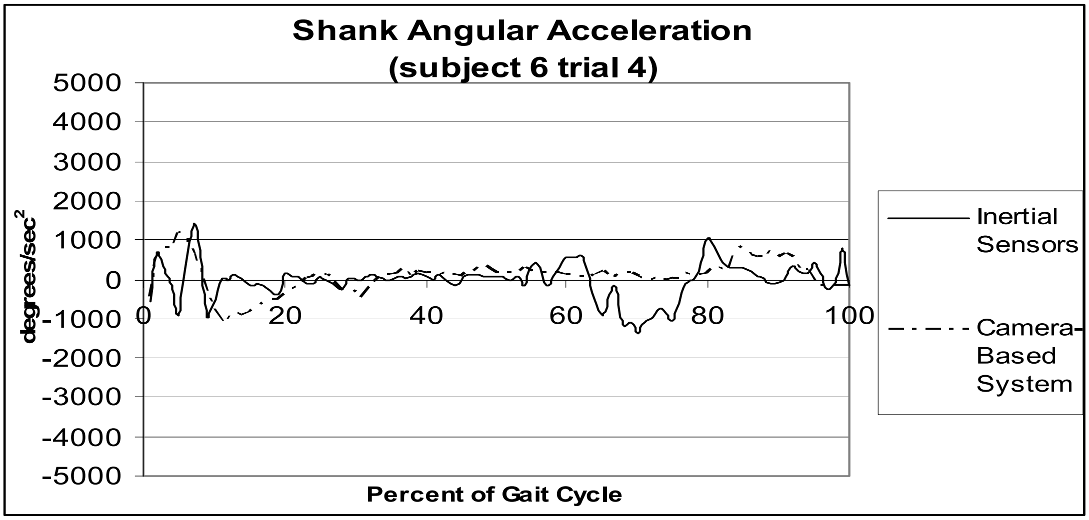

Shank Angular Acceleration
(subject 4 trial 3)
degrees/sec^2
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
-4000
-5000
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

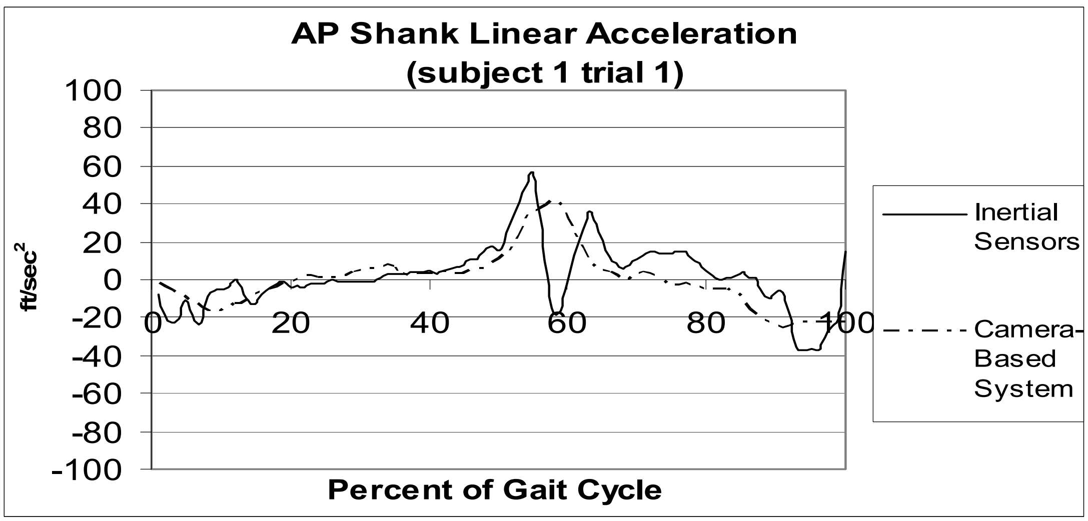

Shank Angular Acceleration
(subject 5 trial 1)
degrees/sec^2
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
-4000
-5000
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

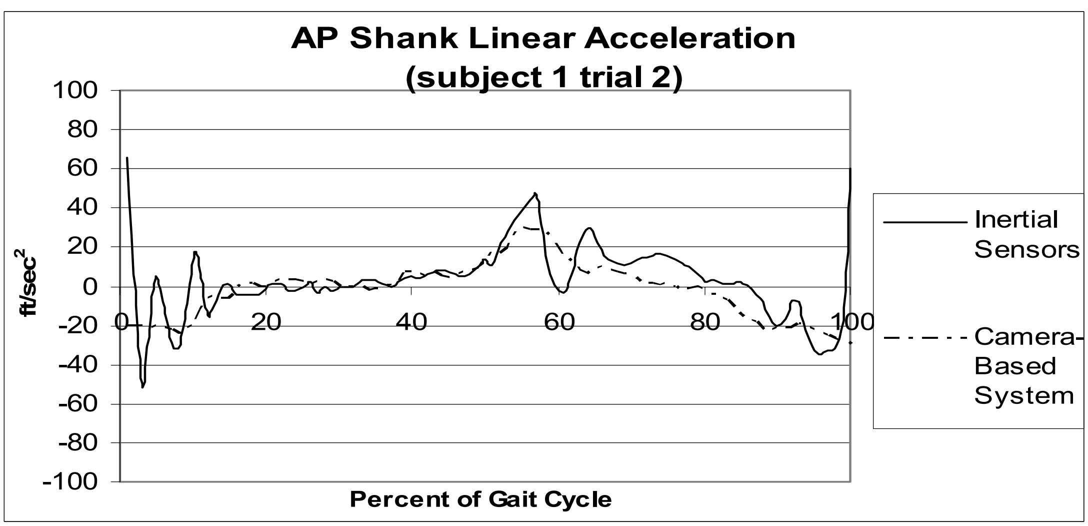

Shank Angular Acceleration
(subject 6 trial 1)
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
-4000
-5000
0
20
40
60
80
100
degrees/sec^2
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

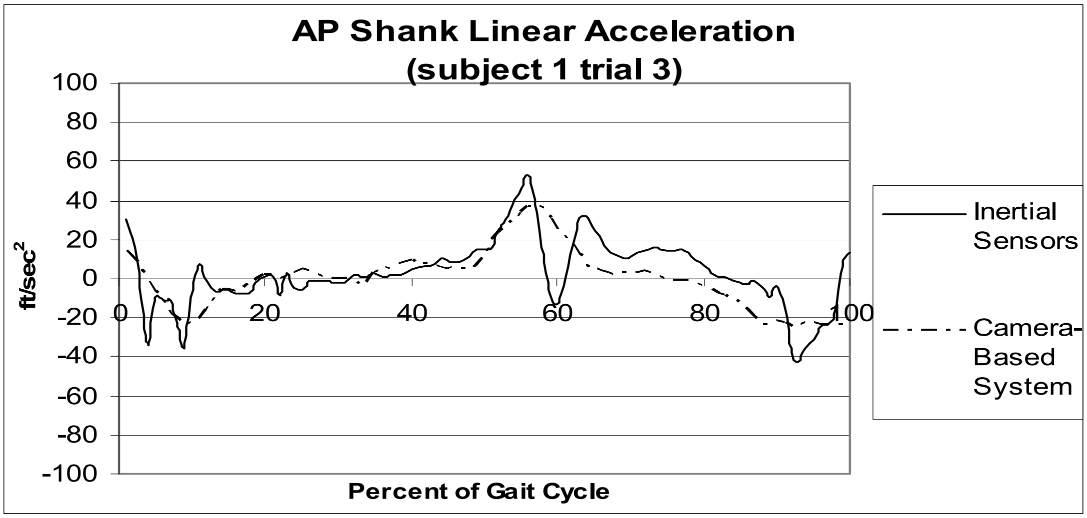

Shank Angular Acceleration
(subject 6 trial 2)
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
-4000
-5000
0
20
40
60
80
100
degrees/sec^2
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

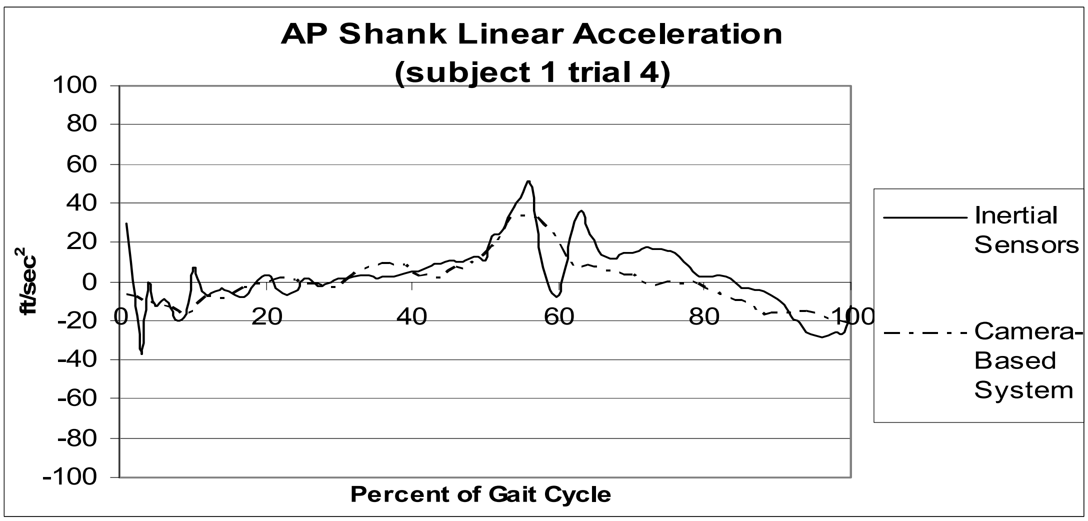

Shank Angular Acceleration
(subject 6 trial 3)
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
-4000
-5000
0
20
40
60
80
100
degrees/sec^2
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

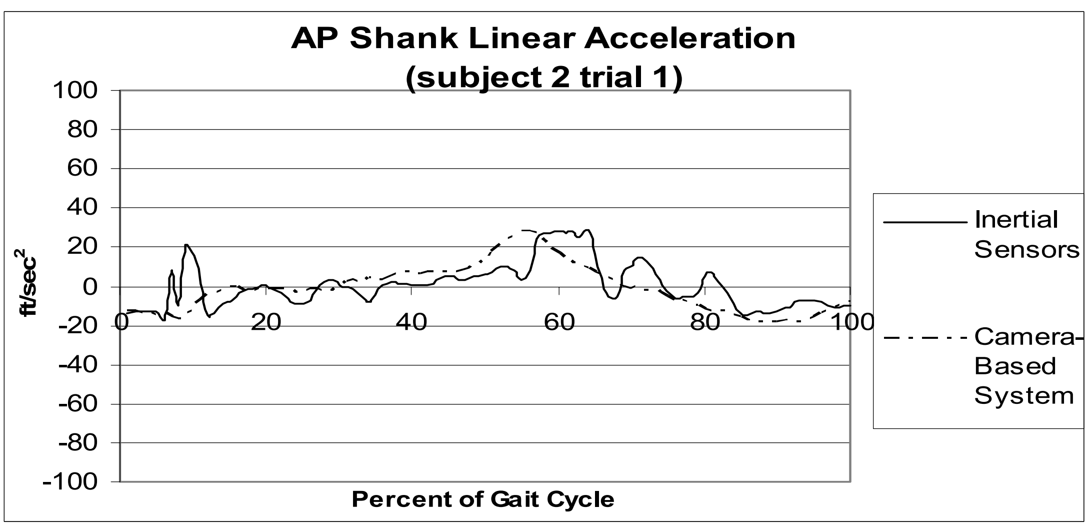

Shank Angular Acceleration
(subject 6 trial 4)
5000
4000
3000
2000
1000
0
-1000
-2000
-3000
-4000
-5000
degrees/sec^2
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

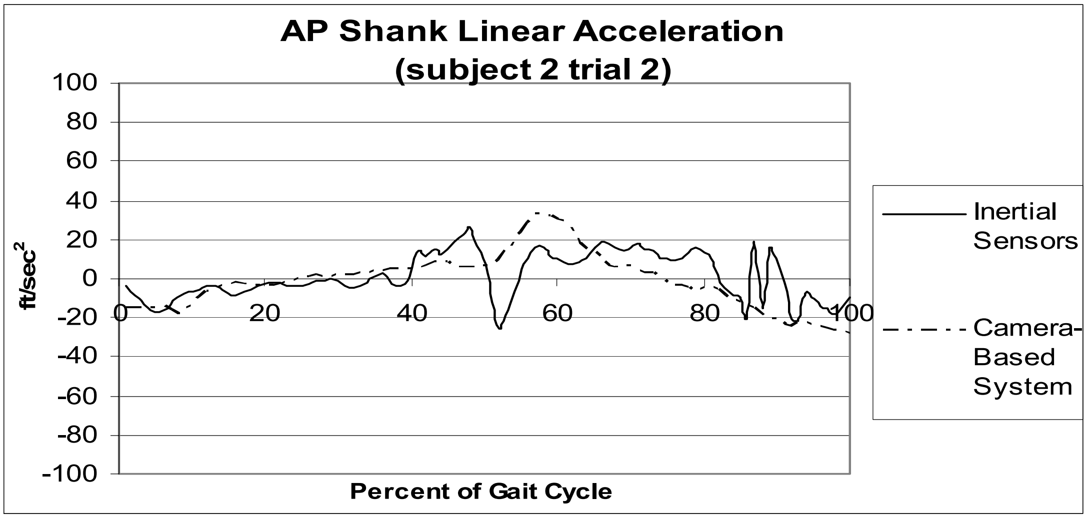

AP Shank Linear Acceleration
(subject 1 trial 1)
100
80
60
40
20
0
-20
-40
-60
-80
-100
ft/sec^2
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

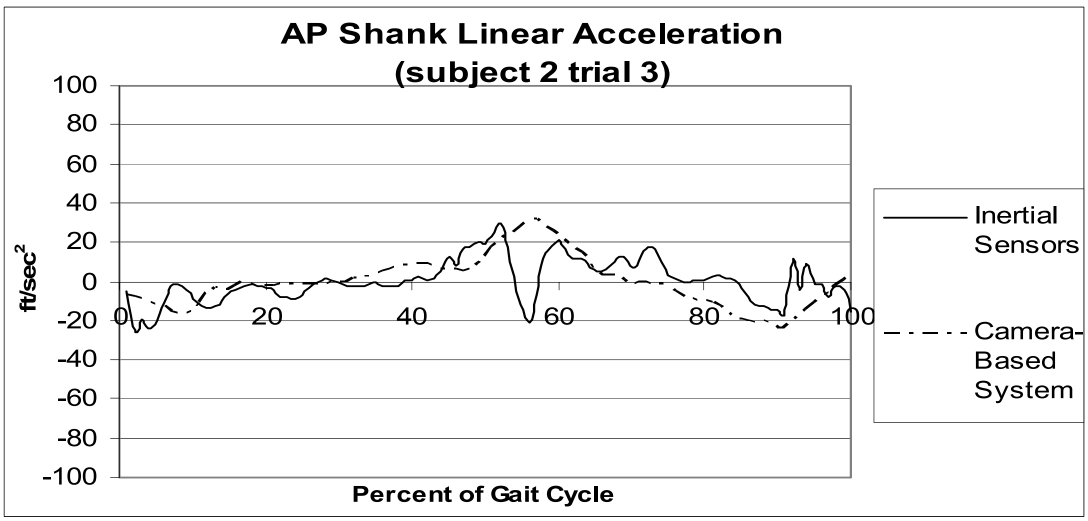

AP Shank Linear Acceleration
(subject 1 trial 2)
100
80
60
40
20
0
-20
-40
-60
-80
-100
ft/sec^2
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

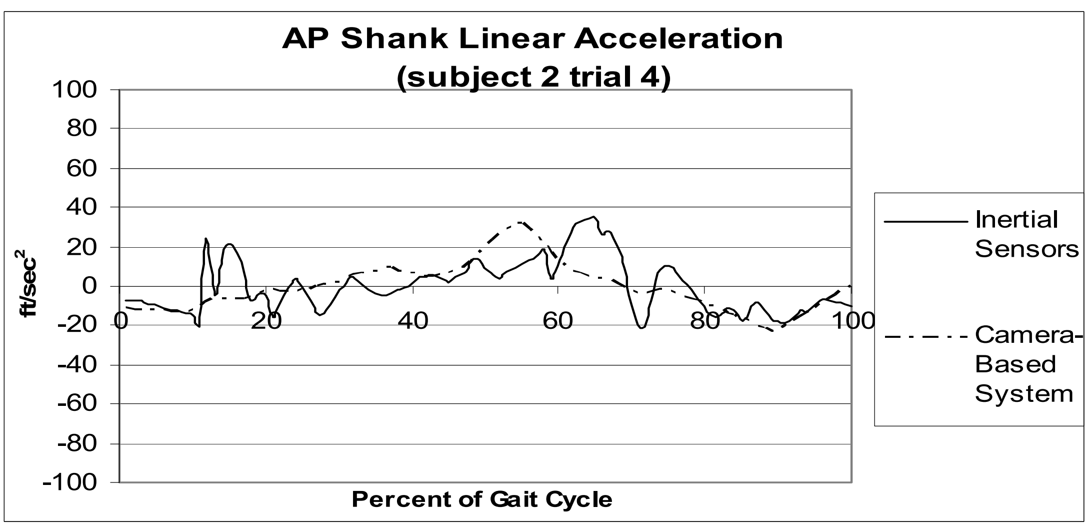

AP Shank Linear Acceleration
(subject 1 trial 3)
ft/sec^2
100
80
60
40
20
0
-20
-40
-60
-80
-100
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

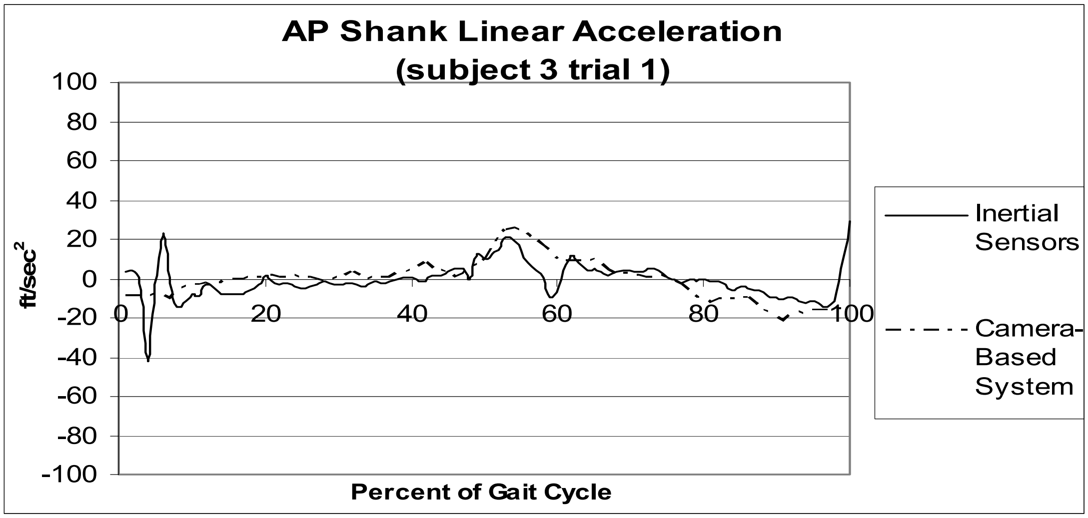

AP Shank Linear Acceleration
(subject 1 trial 4)
ft/sec^2
100
80
60
40
20
0
-20
-40
-60
-80
-100
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

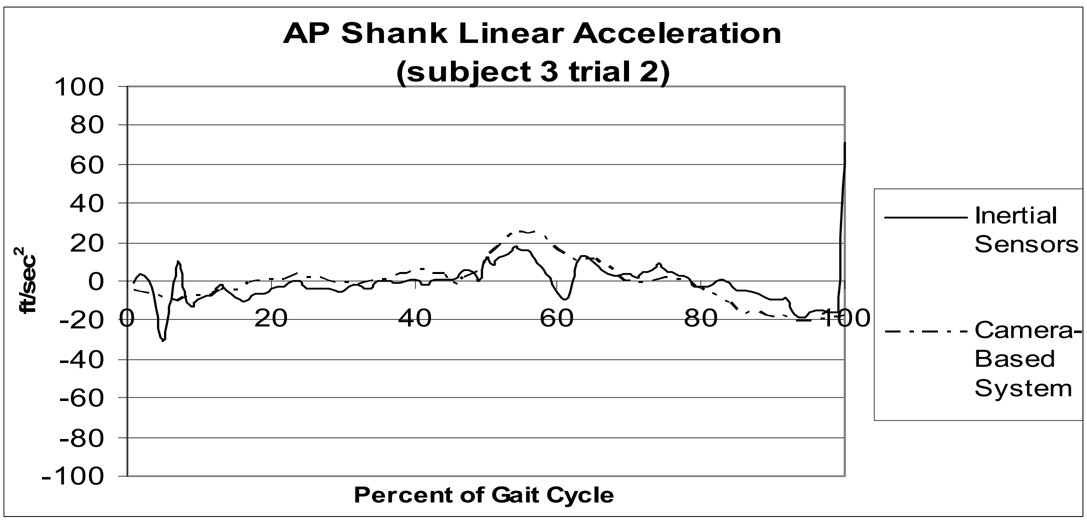

AP Shank Linear Acceleration
(subject 2 trial 1)
ft/sec^2
100
80
60
40
20
0
-20
-40
-60
-80
-100
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

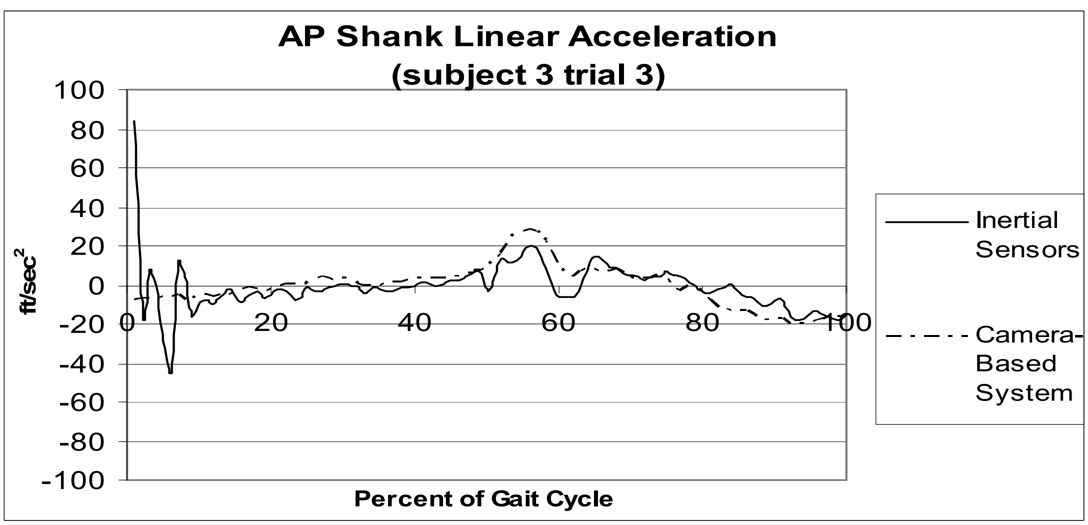

AP Shank Linear Acceleration
(subject 2 trial 2)
ft/sec^2
100
80
60
40
20
0
-20
-40
-60
-80
-100
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

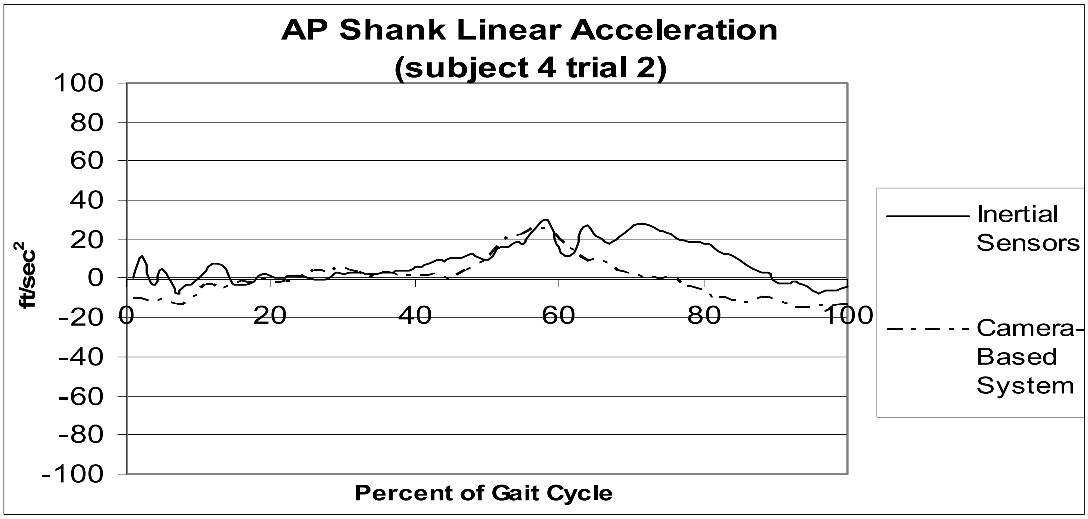

AP Shank Linear Acceleration
(subject 2 trial 3)
ft/sec^2
100
80
60
40
20
0
-20
-40
-60
-80
-100
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

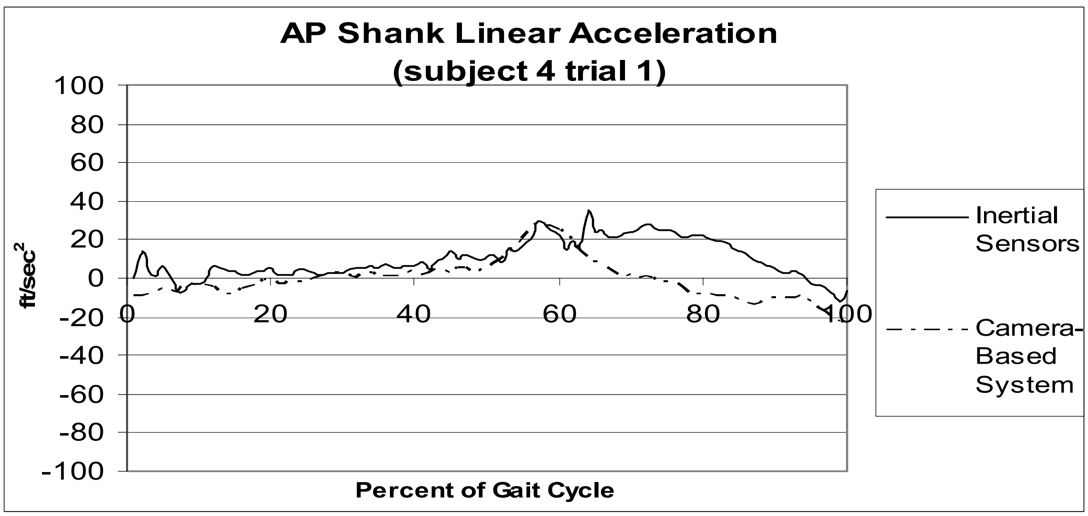

AP Shank Linear Acceleration
(subject 2 trial 4)
ft/sec^2
100
80
60
40
20
0
-20
-40
-60
-80
-100
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

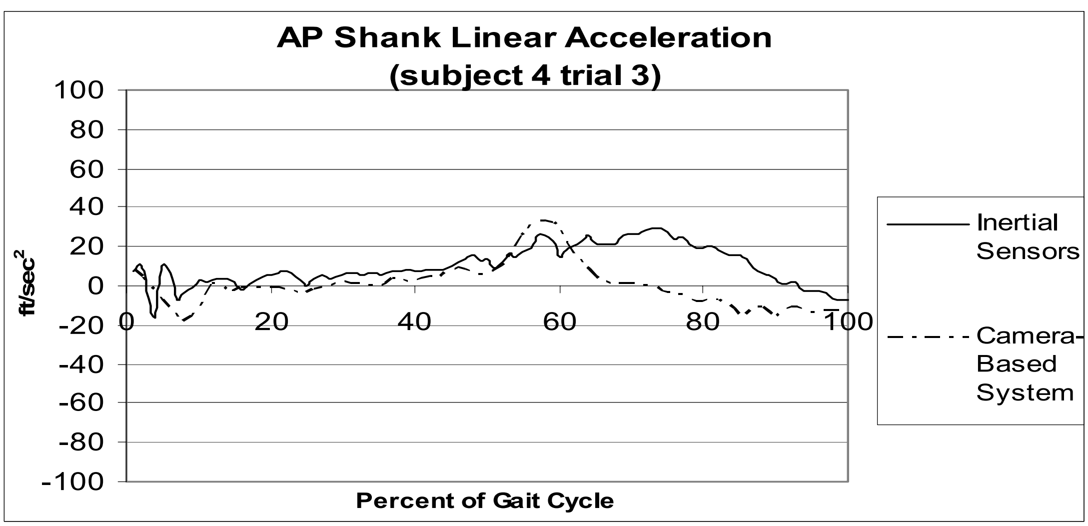

AP Shank Linear Acceleration
(subject 3 trial 1)
100
80
60
40
20
0
-20
-40
-60
-80
-100
ft/sec^2
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

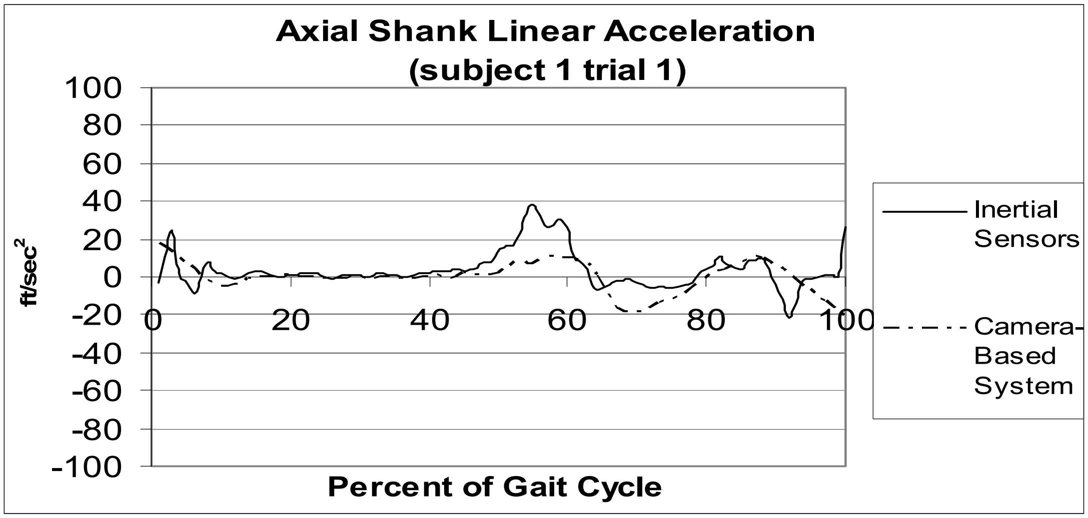

AP Shank Linear Acceleration
(subject 3 trial 2)
100
80
60
40
20
0
-20
-40
-60
-80
-100
ft/sec^2
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

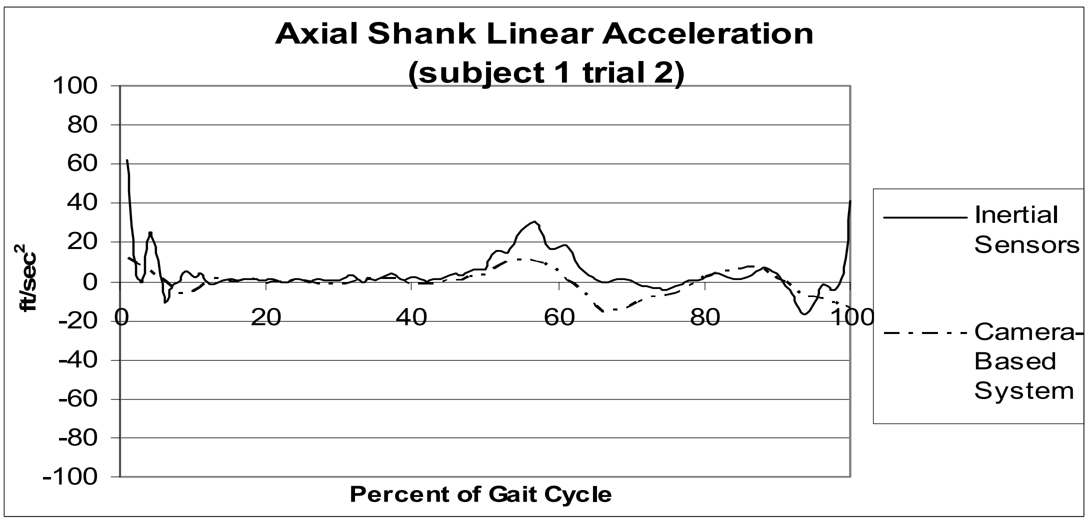

AP Shank Linear Acceleration
(subject 3 trial 3)
100
80
60
40
20
0
-20
-40
-60
-80
-100
ft/sec^2
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

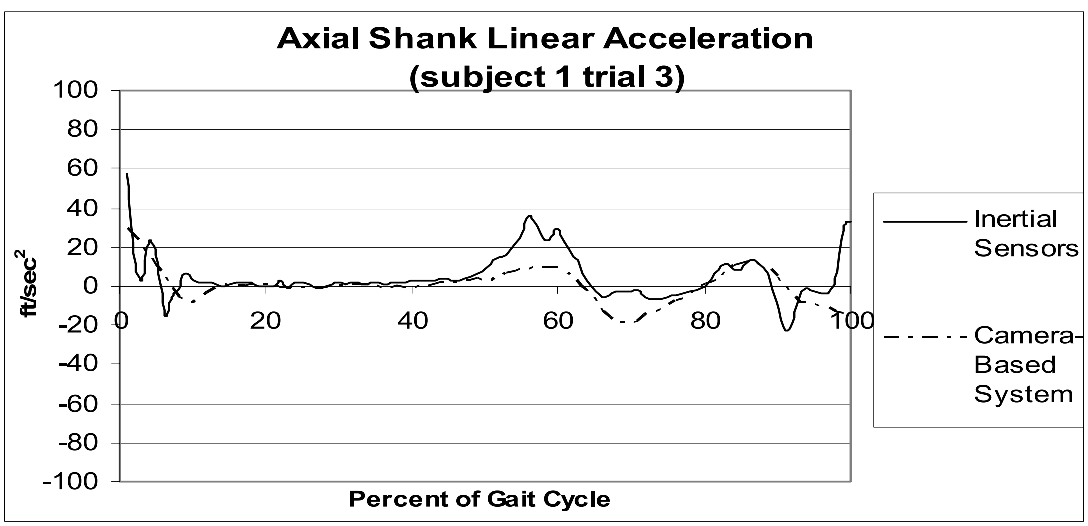

AP Shank Linear Acceleration
(subject 4 trial 2)
ft/sec^2
100
80
60
40
20
0
-20
-40
-60
-80
-100
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

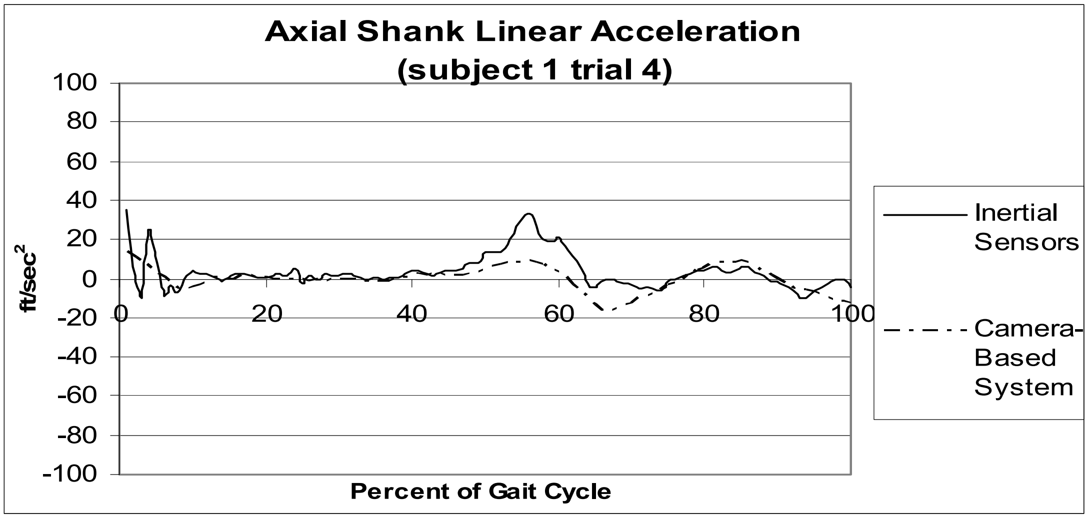

AP Shank Linear Acceleration
(subject 4 trial 1)
ft/sec^2
100
80
60
40
20
0
-20
-40
-60
-80
-100
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

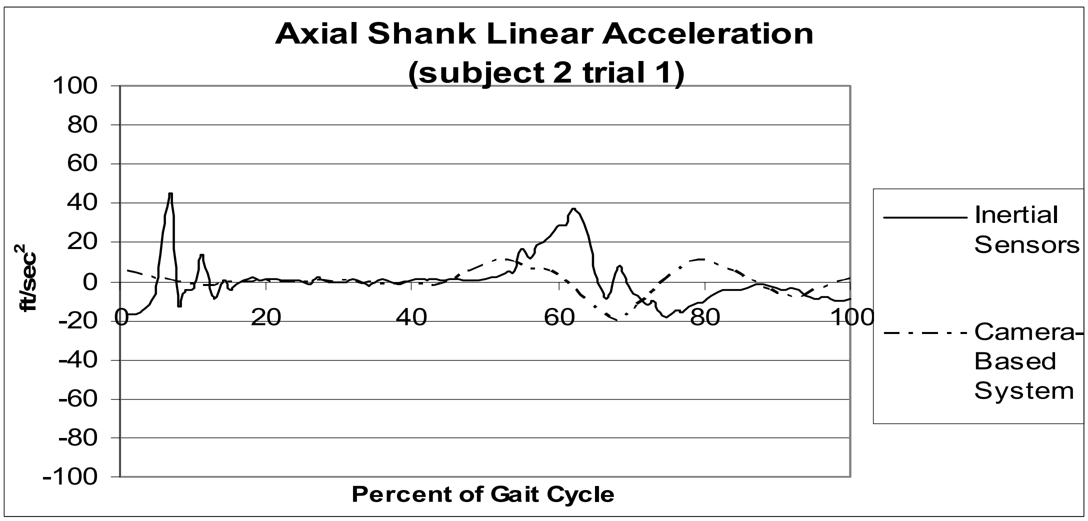

AP Shank Linear Acceleration
(subject 4 trial 3)
ft/sec^2
100
80
60
40
20
0
-20
-40
-60
-80
-100
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

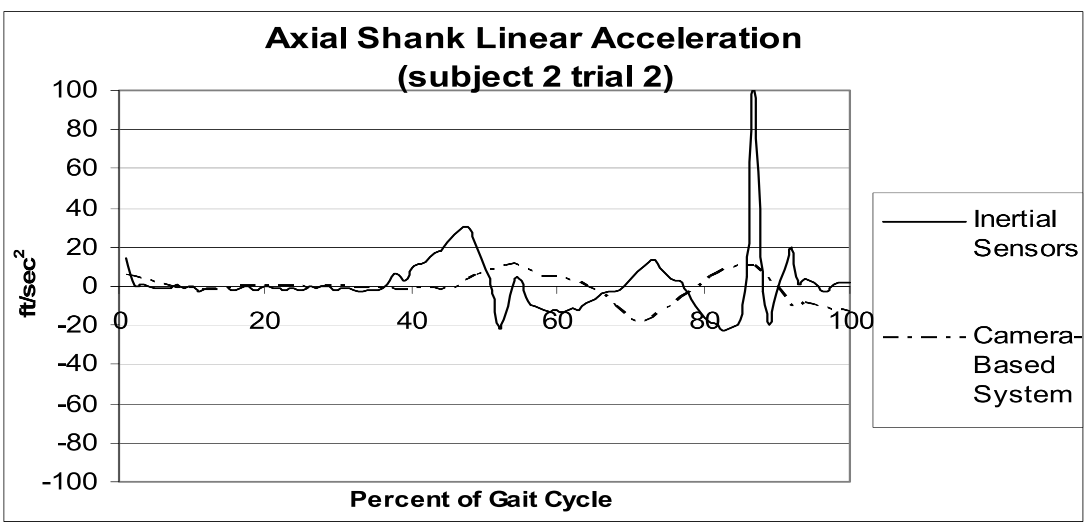

Axial Shank Linear Acceleration
(subject 1 trial 1)
100
80
60
40
20
0
-20
-40
-60
-80
-100
ft/sec^2
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

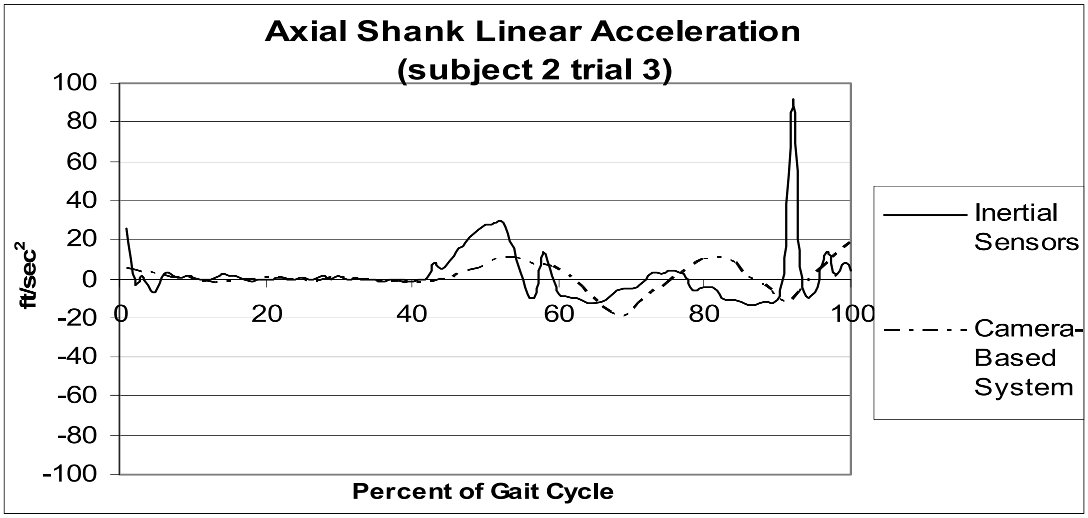

Axial Shank Linear Acceleration
(subject 1 trial 2)
100
80
60
40
20
0
-20
-40
-60
-80
-100
ft/sec^2
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

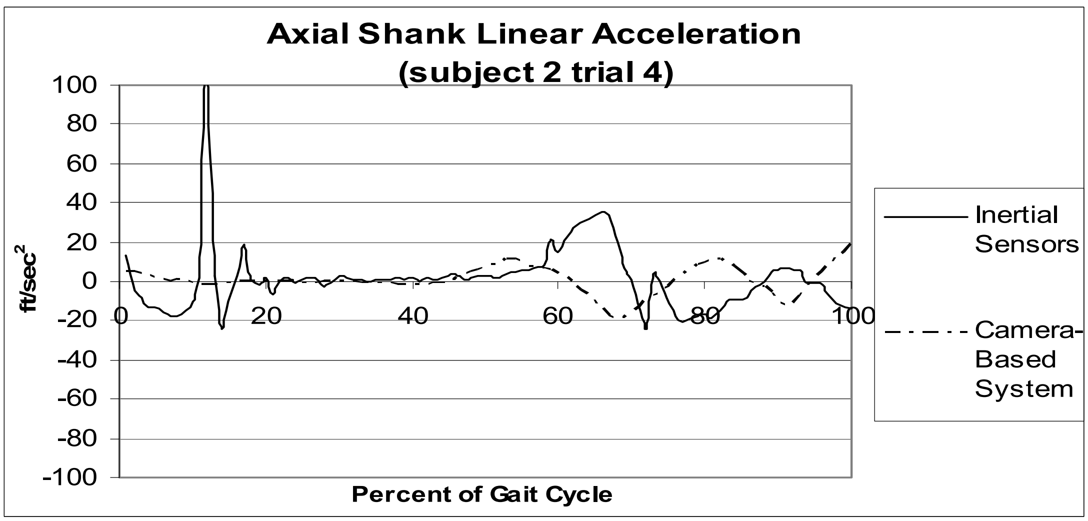

Axial Shank Linear Acceleration
(subject 1 trial 3)
100
80
60
40
20
0
-20
-40
-60
-80
-100
ft/sec^2
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

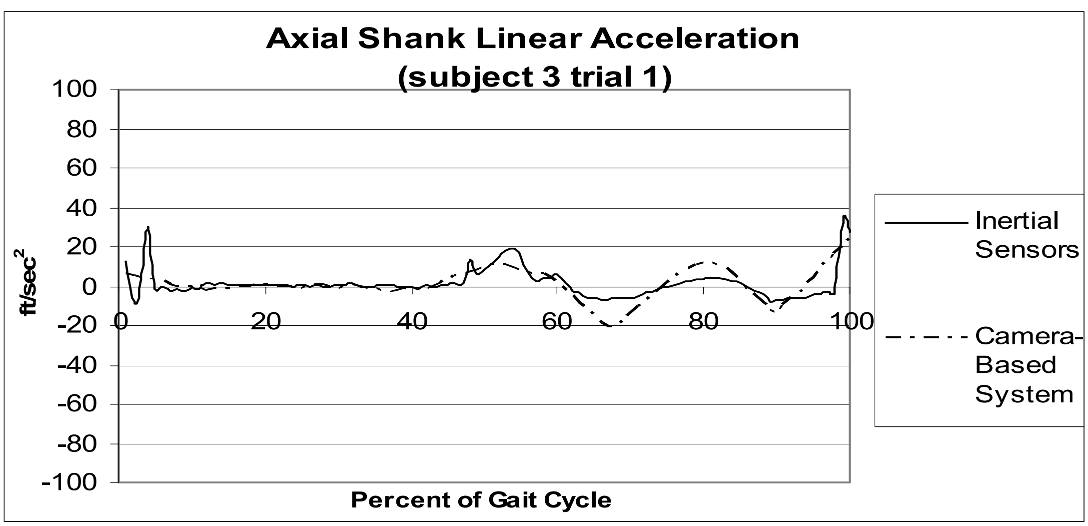

Axial Shank Linear Acceleration
(subject 1 trial 4)
ft/sec^2
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

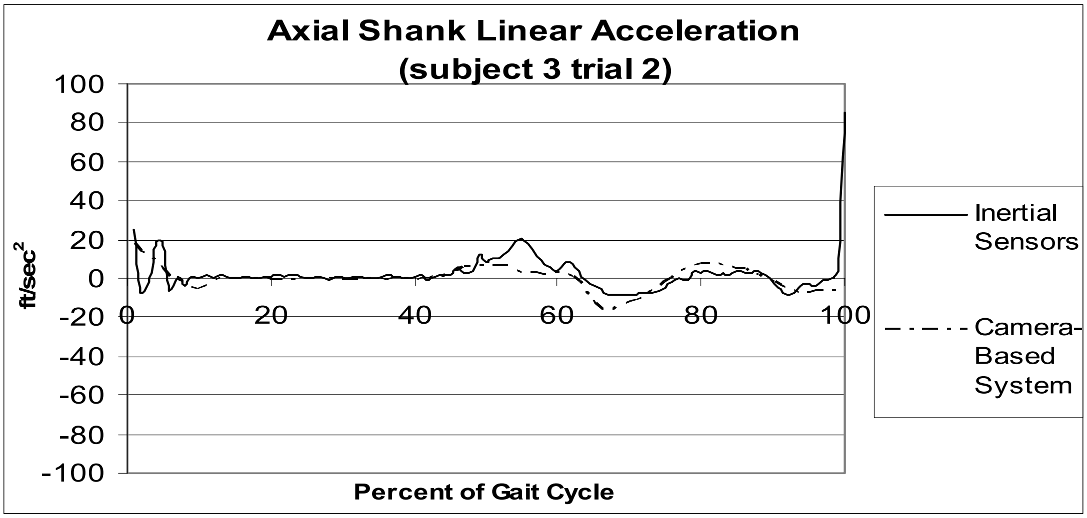

Axial Shank Linear Acceleration
(subject 2 trial 1)
ft/sec^2
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

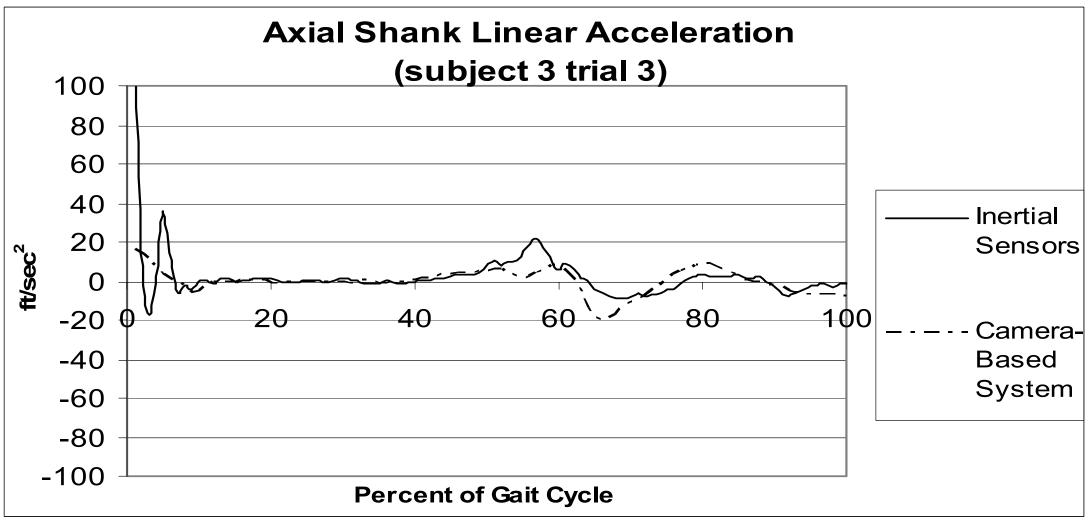

Axial Shank Linear Acceleration
(subject 2 trial 2)
ft/sec^2
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

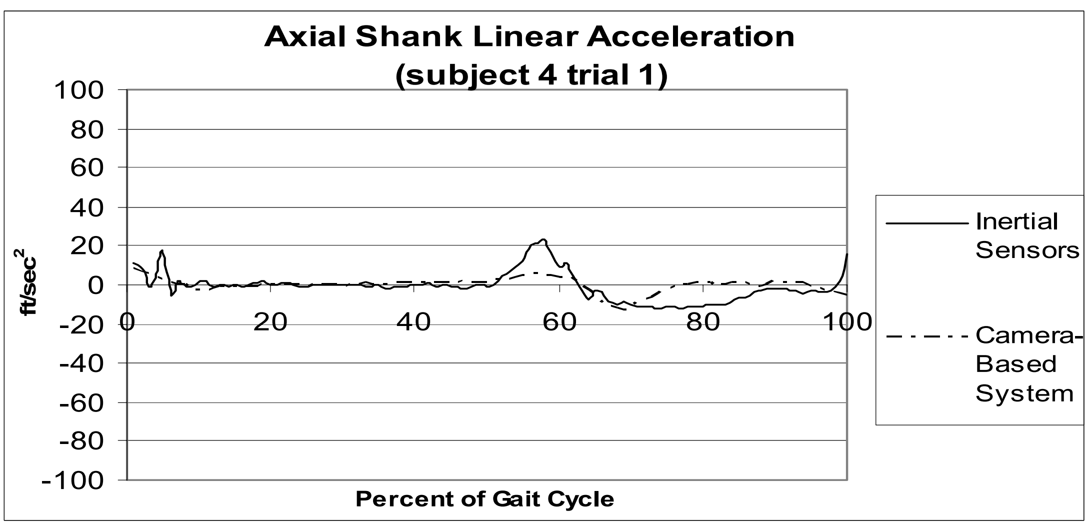

Axial Shank Linear Acceleration
(subject 2 trial 3)
ft/sec2
100
80
60
40
20
0
-20
-40
-60
-80
-100
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

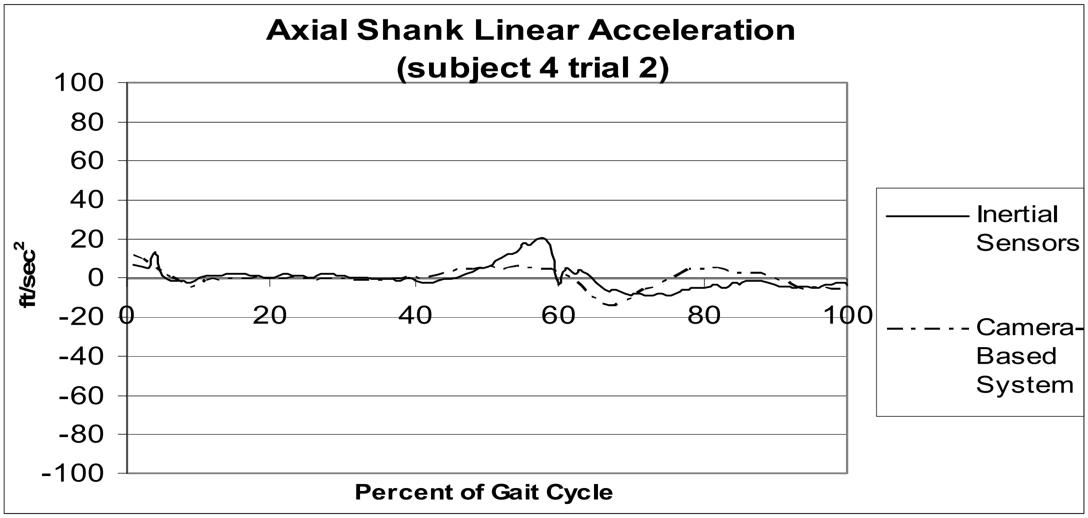

Axial Shank Linear Acceleration
(subject 2 trial 4)
ft/sec2
100
80
60
40
20
0
-20
-40
-60
-80
-100
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System


Axial Shank Linear Acceleration
(subject 3 trial 1)
ft/sec2
100
80
60
40
20
0
-20
-40
-60
-80
-100
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

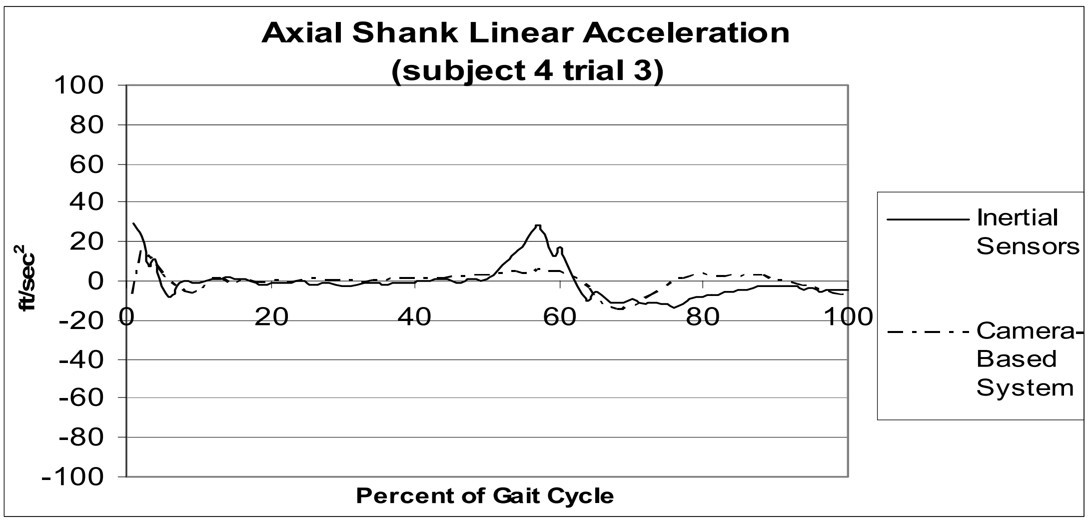

Axial Shank Linear Acceleration
(subject 3 trial 2)
100
80
60
40
20
0
-20
-40
-60
-80
-100
0
20
40
60
80
100
ft/sec^2
Percent of Gait Cycle
Inertial Sensors
Camera-Based System


Axial Shank Linear Acceleration
(subject 3 trial 3)
100
80
60
40
20
0
-20
-40
-60
-80
-100
0
20
40
60
80
100
ft/sec^2
Percent of Gait Cycle
Inertial Sensors
Camera-Based System


Axial Shank Linear Acceleration
(subject 4 trial 1)
100
80
60
40
20
0
-20
-40
-60
-80
-100
0
20
40
60
80
100
ft/sec^2
Percent of Gait Cycle
Inertial Sensors
Camera-Based System



Axial Shank Linear Acceleration
(subject 4 trial 2)
100
80
60
40
20
0
-20
-40
-60
-80
-100
ft/sec²
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System



Axial Shank Linear Acceleration
(subject 4 trial 3)
100
80
60
40
20
0
-20
-40
-60
-80
-100
ft/sec²
0
20
40
60
80
100
Percent of Gait Cycle
Inertial Sensors
Camera-Based System

Appendix 2

Steps of Data Processing

1) Data Collection and Processing
   a) Data was collected and saved using Labview®
      i) Labview uses the pre-trial calibration values to produce 2 files
         (1) One with the raw voltages collected
            (a) Designated by the date, the subject's initials, and the trial number with the suffix "raw" in the form of mmddyyiii#_raw
         (2) The second file created uses the calibration data to produce values for the measured parameters
            (a) Designated by files named mmddyyiii#
      ii) The mmddyyiii# files were then input into a Matlab 7.0® program to perform the necessary integration and differentiation of the angular velocity data
         (1) A command in the Matlab program saves the data and calculated results into an Microsoft Excel® worksheet
      iii) The values from the worksheet were then pasted into a template worksheet in Excel for further manipulation
         (1) The processed file was saved in the form mmddyyiii#_proc

b) For each subject 3 files were made using Labview during the calibration steps with a different one of the 3 axes of the accelerometer being perpendicular to the ground during each file

i) Files were saved as before as a raw voltage file and calibrated file with the designations mmddyyiii(x, y, or z) and mmddyyiii(x, y, or z)_raw

(1) The perpendicular axis of the accelerometer designates whether the step is saved with the x, y or z suffix

ii) Each of the 3 files was then pasted into the Excel worksheet template called "orientation template left" or "orientation template right" depending on the leg that was tested

(1) The template contained specific columns for the acceleration files depending on the axis perpendicular to the ground

(2) The values from the 3 files are used to calculate the cosines needed to transform the accelerometers axis to a shank based coordinate system

(a) A sheet in the template contains the transformation matrix for the conversion

iii) The transformation matrix from the orientation template is pasted into the appropriate cells in the mmddyyiii#_proc file for each of the subjects trials

2) Correcting the data

a) Drift in the angular velocity data of G1 and G3 must be corrected to prevent propagation of the error to other parameters

   i) Since the recorded data begins and ends at rest the angular velocity should end at 0˚/sec

   ii) Using the angular velocity values of G1 and G3 from mmddyyiii#_proc the rate of the drift is calculated

      (1) Rate of drift [D] = terminal angular velocity value at rest [AV_T]/ (the time when terminal angular velocity becomes level [T_T]– the time when initial angular velocity changes from 0˚/sec [T_I])

   iii) The rate of drift is then used to subtract the error from the effected data

      (1) Angular velocity(new) = Angular velocity(old) – D*(t – T_I)

         (a) from $t = T_I$ to $t = T_T$

   iv) The corrected angular velocities are then pasted into the mmddyyiii# file to replace the previous values and saved as a text file

b) The mmddyyiii#.txt file is run through the Matlab program to correct the calculations

   i) The output file name was changed before running the program to prevent duplicate naming

   ii) The worksheet values are then pasted over the values in ddmmyyiii#_proc

(1) File → Save As file name: "ddmmyyiii#_corrected"

(2) This was done to maintain both copies of the file

c) ddmmyyiii#_corrected will have the corrected values for G1 and G3 as well as all calculations using those points

3) Comparing data

a) The trials for both systems were pasted into a new worksheet specific for that parameter

b) The systems collect data points at different frequencies so the single gait cycle was splined using Excel

i) Each systems entire gait cycle was converted to 100 points

c) Comparison then proceeded by comparing points at a specific percentage of the gait cycle instead of at time intervals