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Reliable Sagittal Plane Kinematic Gait Assessments are Feasible using Low-Cost Webcam Technology

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Abstract

Three-dimensional (3-D) motion capture systems are commonly used for gait analysis because they provide reliable and accurate measurements. However, the downside of this approach is that it is expensive and requires technical expertise; thus making it less feasible in the clinic. To address this limitation, we recently developed and validated (using a high-precision walking robot) a low-cost, two-dimensional (2-D) real-time motion tracking approach using a simple webcam and LabVIEW Vision Assistant. The purpose of this study was to establish the repeatability and minimal detectable change values of hip and knee sagittal plane gait kinematics recorded using this system. Twenty-one healthy subjects underwent two kinematic assessments while walking on a treadmill at a range of gait velocities. Intraclass correlation coefficients (ICC) and minimal detectable change (MDC) values were calculated for commonly used hip and knee kinematic parameters to demonstrate the reliability of the system. Additionally, Bland-Altman plots were generated to examine the agreement between the measurements recorded on two different days. The system demonstrated good to excellent reliability (ICC > 0.75) for all the gait parameters tested on this study. The MDC values were typically low (< 5°) for most of the parameters. The Bland-Altman plots indicated that there was no systematic error or bias in kinematic measurements and showed good agreement between measurements obtained on two different days. These results indicate that kinematic gait assessments using webcam technology can be reliably used for clinical and research purposes.

Keywords

Biomechanics; Kinematic analysis; Gait tracking; Real-time; Camera systems

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INTRODUCTION

Computerized gait assessment using a three-dimensional (3-D) camera system is considered to be the ideal approach for kinematic gait analysis. A major advantage of 3-D systems is that their measurements are highly accurate and reliable [1]. However, the downside of this technology is that it is expensive, thus making it unaffordable to many clinicians and researchers [2], particularly in developing countries. As a result, clinicians often rely on qualitative gait analysis to document patient progress during rehabilitation. However, the outcomes from qualitative analysis are subjective and often lack sufficient reliability [3–5]. Thus, the issue of establishing a cost-effective, reliable method of gait tracking needs to be addressed.

Along these lines, previous studies have worked on several marker-based and markerless techniques (using a single video camera) to perform quantitative gait analysis [6–9]. While these approaches simplify experimental setup and typically provide valid and reliable kinematic data [6, 8], they are often limited to off-line analysis of gait kinematics; thus, making it labor intensive during the processing phase. Further, systems that require manual identification of anatomical landmarks to process kinematic data lack sufficient reliability due to digitization errors [6, 7]. To address these issues, we recently developed a low-cost, real-time motion tracking system for quantitative gait analysis and validated the outcomes using a high-precision robotic gait device [10]. This system uses a simple webcam and an image processing algorithm to track the thigh and leg segments in real-time and compute hip and knee kinematics. The system is portable, quick to set-up (~5 minutes), and has a graphical user interface that makes it easy to operate in comparison with other low-cost systems. The system also provides online feedback of the subject's gait parameters, which could be particularly useful for rehabilitation of gait function after neurological impairment [10–12].

Although this low-cost system has shown to be accurate in measuring gait parameters, repeatability of the system is still unknown. A clear understanding of the reliability of the webcam system is crucial, as poor repeatability can affect outcomes of evaluation and clinical-decision making process; thus making it unsuitable for practical purposes. Accordingly, this study evaluated the repeatability of the low-cost, 2-D kinematic webcam system and established the minimal detectable change scores for several commonly used kinematic gait parameters obtained using this system.

METHODS

Subjects

Twenty-one healthy adults (14 males, 7 females; 23.24 ± 6.33 years; 1.75 ± 0.01 meters; 74.24 ± 17.85 kilograms) participated in this study over two testing sessions that were separated by about one week (5.19 ± 4.67 days). All subjects signed an informed consent document that was approved by the Institutional Review Board of the University of Michigan.

Kinematic Tracking using Webcam Technology

The details of the hardware and software for real-time kinematic tracking are provided in detail elsewhere [10]. Briefly, the system consists of a Logitech-C920 HD Pro Webcam (1080p, 30 FPS), a SunPAK 6600DX heavy duty tripod, a Rigid Industries floodlight, and standard 19 mm retroreflective markers. The marker data representing the thigh and shank segments were collected and processed in real-time using a custom image processing algorithm written in LabVIEW Vision Assistant [10] (National Instruments Corp., Austin, TX, USA). Images were processed as follows: Initially, at startup, the program captures a frame of the video using IMAQdx Grab VI [Video Mode=32 (800×600 MJPG 30FPS), Gain=255 (maximum), Exposure=0 (minimum); Brightness=128 (medium); and Contrast=1.28 (medium)]. A region of interest is then selected using the IMAQ ConstructROI VI, so as to analyze only the area where the subject would be walking. The program then filters the image data for the brightest (whitest) parts of the image using the IMAQ IVA ColorThreshold VI and Binary Inverse VI. Here, pixels past a certain whiteness threshold are set to 1 (i.e., all with RGB values greater than [255, 255, and 234]), the rest are set to 0. Thus, all pixels containing markers are set to 1. Next, the program discards everything outside the region of interest using the IMAQ IVA MaskfromROI VI. The program looks at pixel values of 1 and selects items that are in the shape of a circle within a specified radius using IMAQ Find Circles VI. Lastly, the program outputs pixel coordinates of all these circles (i.e., the markers), which are provided in real-time. A three-point kinematic model was then created from the hip, knee, and ankle markers and was used to compute the sagittal plane hip and knee kinematics.

Experimental Procedures

The experiment was performed in a regular biomechanical laboratory environment. Adequate measures (blackout blinds, background screen, and non-reflective clothing) were taken to minimize spurious detection of extraneous objects as markers and to optimize tracking performance. Retroreflective markers (19mm) were then placed on the subject's greater trochanter, lateral femoral epicondyle, and lateral malleolus while they stood on the center of a motorized treadmill (Woodway USA Inc.). The camera was positioned such that the center of the camera (optical axis) was aligned with the lateral aspect of the subject's trunk (i.e., in-line with the center of the treadmill where the center part of the motion is expected to occur while walking) – this ensured that the parallax error was minimal during the experiment. After which, the region of interest for data capture was selected using the data collection software and the brightness, exposure and gain of the camera were adjusted to maximize tracking performance. The subject then walked over a motorized treadmill at seven different speeds (1.0, 1.5, 2.0, 2.5, 3.0, 3.5, and 4.0 mph) with their hands placed on a custom built treadmill rail system. The order of walking speeds was pseudo-randomized (beginning at 2.0 mph and ending at 1.0 mph) across subjects. At the beginning of each trial, the subject was adapted to the walking speed for 30 seconds. Following which, kinematic data were collected for one minute while the subject walked over the treadmill. After completing all testing trials, the subjects returned to the laboratory within one week of the initial testing, and repeat testing was performed in an identical manner.

Data Processing

The hip and knee angle data were segmented from each heel strike to the subsequent heel strike, interpolated to 100 points, and then ensemble averaged to give a single trace for each trial. Using these traces, we calculated the variables of interest used in this study. For the hip we calculated excursion over the entire gait cycle, peak extension for the entire gait cycle, peak flexion during the stance phase, and peak flexion during the swing phase. For the knee we calculated the excursion for the entire gait cycle, peak extension during the stance phase, peak flexion during the stance phase, peak flexion during the swing phase, and excursion during the stance phase. The stance and swing phases were determined using two methods. The first method used a simple clinician-friendly approach where the stance phase was considered to be the first 60% (or 60 data points) of the gait cycle while the swing phase occupied the remaining 40% [13–16]. The second method used the horizontal ankle position (in relation to the hip) derived using a forward kinematic approach to determine the stance and swing phases of the gait cycle [10, 17] (Supplementary Material). The repeatability of knee and hip variables obtained using both these methods was evaluated.

Statistical Analysis

All data analyses were performed in SPSS for windows version 22 (SPSS Inc., Chicago, IL, USA) and R statistical software (version 3.1.3). ICC values were calculated using a two-way mixed effects model for single measurement and absolute agreement [i.e., ICC(3,1) model]. MDC scores were calculated using the following equations [18].

$$\text{SEM} = \text{SD} \times \sqrt{1 - \text{ICC}} \quad (1)$$

$$\text{MDC} = \text{SEM} \times 1.96 \times \sqrt{2} \quad (2)$$

In these equations, SEM is the standard error of measurement and SD is the average of the standard deviations for sessions 1 and 2 of the study. Bland-Altman plots were used to examine the limits of agreement between the measurements recorded on the two testing sessions.

RESULTS

The hip and knee kinematic profiles obtained on two different days were almost identical to each other (Figure 1). ICC and MDC values are provided in Table 1 and Supplementary Table 1. Average values of the sagittal plane hip and knee kinematics are provided in Supplementary Tables 2 and 3. The hip and knee kinematic gait parameters showed good to excellent reliability for all tested velocities (ICC = 0.75 to 0.93) when using the 60:40 (stance-to-swing ratio) approach, but lacked sufficient reliability for stance phase knee excursion at 1 mph when using the actual toe-off events calculated using the forward kinematic approach (Supplementary Table 1). The MDC values typically ranged between 3° to 5° for the hip parameters and between 4° to 8° for the knee parameters (Table 1 and

Supplementary Table 1). Bland-Altman plots showed good agreement between measurements recorded on two different days and demonstrated no systematic bias in test-retest errors for the hip and knee kinematic parameters (Figure 2 and Supplementary Figure 1).

DISCUSSION

This study tested the reliability of a low-cost camera system designed for 2-D gait analysis [10]. The results indicated that kinematic gait measurements obtained from the low-cost system were reliable, as evidenced by high ICC values and low MDC scores. Additionally, Bland-Altman plots showed good agreement between test-retest measurements for all kinematic parameters. For these reasons, the gait analysis system tested in this study could add utility in clinics as well as in research; especially in the developing world, where cost is prohibitive to the use of sophisticated gait analysis systems.

The reliability of motion capture systems is typically evaluated at self-selected walking speed [1, 19–21]. This is understandable considering that kinematic assessments for clinical and research purposes are often performed at self-selected walking speed. However, in some cases (e.g., energy expenditure research), kinematic assessments are performed over a range of walking speeds. Thus, to get a complete picture of the system's utility, we computed ICC and MDC scores across a spectrum of gait speeds. Interestingly, we found that the reliability of the system was not affected by the subject's gait speed. Further, the reliability coefficients and MDC scores observed are comparable to those obtained from 3D camera systems (Supplementary Table 4) [1]. Similarly, the observed changes in gait kinematic profiles with increasing walking speed (Supplementary Figure 2) were also consistent with previous studies that have used sophisticated 3D motion capture systems [20, 22–24].

Accurate and reliable detection of ground contact/toe-off events is critical for gait analysis. Conventionally, gait events are detected using an instrumented treadmill or embedded force plates on the ground. However, such systems are expensive and are thus not clinically feasible. To simplify the process of gait event detection, the 60 and 40 percent division of the gait cycle was applied as this approach is more likely to be used by clinicians. Although this is a simple procedure, it does not always yield similar results to those using actual gait events across all gait speeds and subjects when detecting toe-off. In order to achieve better accuracy, the trajectory of the horizontal ankle position (x-coordinate of the ankle) in relation to the hip was determined using a forward kinematic model to detect gait events [17, 25, 26] and the results were compared to the data obtained using a force plate treadmill (Supplementary Materials). Similar to other studies that have used 3-D camera systems [17, 25, 26], we found that the horizontal ankle position during gait could serve as an accurate surrogate method to define stance and swing phases of gait cycles (Supplementary Table 5).

These findings indicate that reliable sagittal plane kinematic measurements are feasible using low-cost webcams. Further, the clinical utility of webcam-based motion tracking is high as only minimal expertise is required to operate this system. The real-time processing of kinematic data also allows for the provision of kinematic biofeedback, which could be

useful in facilitating therapeutic outcomes [11, 12]. For these reasons, webcam-based motion tracking can serve as a suitable alternative when 3D systems are unavailable.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

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- Three-dimensional motion analysis systems are expensive for typical clinical use
- Here, we tested the reliability of a webcam-based kinematic gait analysis system
- We found that the system is reliable for assessing sagittal plane gait kinematics
- Reliability scores were similar to those of three-dimensional camera systems
- Reliability of the developed system is appropriate for clinical and research use

Figure 1.

Plots showing the repeatability of sagittal plane hip and knee kinematic profiles across the seven tested speeds. Kinematic profiles were ensemble averaged across strides and subjects. Solid and dotted lines represent the profiles captured on the first (Day 1) and second (Day 2) testing session, respectively. The X-axes represent the percentage of the gait cycle (0–100%) and the Y-axes represent the hip and knee angles in degrees. Note that the kinematic profiles obtained on two testing sessions are almost identical. Detailed interpretation of the values on the Y-axes can be seen in Supplementary Figure 2.

Figure 2.

Bland-Altman plots showing the agreement in hip and knee kinematic gait parameters between the two testing sessions. The X-axes represent the average of measurements obtained on the two testing sessions and the Y-axes represent the differences in measurements (i.e., error) between the two testing sessions. Note: For clarity purposes, Bland-Altman plots are shown only for 2.5 mph.

Table 1

Repeatability measures of the low-cost 2D webcam system.

ICC		1.0	1.5	2.0	2.5	3.0	3.5	4.0
		mph	mph	mph	mph	mph	mph	mph
Hip	Excursion	0.88	0.82	0.83	0.84	0.89	0.92	0.81
	Peak Extension	0.76	0.76	0.75	0.80	0.84	0.77	0.83
	Peak Flexion Stance	0.85	0.90	0.91	0.88	0.88	0.88	0.79
	Peak Flexion Swing	0.89	0.87	0.87	0.85	0.85	0.84	0.81
Knee	Excursion	0.78	0.85	0.88	0.93	0.90	0.93	0.85
	Peak Extension Stance	0.84	0.86	0.85	0.83	0.77	0.77	0.85
	Peak Flexion Stance	0.84	0.85	0.77	0.86	0.82	0.76	0.89
	Peak Flexion Swing	0.86	0.89	0.88	0.91	0.90	0.90	0.86
MDC	Stance Excursion	0.84	0.86	0.91	0.90	0.93	0.89	0.91
	Excursion	3.5° (10.6)	4.8° (14.1)	3.4° (9.31)	3.4° (8.87)	3.0° (7.41)	2.6° (6.20)	5.1° (11.9)
	Peak Extension	4.9° (37.4)	5.5° (42.2)	5.1° (37.8)	4.5° (30.2)	4.1° (25.5)	4.5° (26.2)	4.0° (21.7)
	Peak Flexion Stance	4.7° (31.9)	3.4° (19.8)	3.2° (17.5)	4.1° (20.6)	3.7° (17.5)	3.6° (16.5)	4.9° (21.2)
Hip	Peak Flexion Swing	3.6° (18.0)	3.7° (17.5)	3.6° (15.7)	3.8° (15.9)	4.1° (16.8)	4.0° (16.5)	4.4° (17.7)
	Excursion	5.4° (9.16)	5.4° (8.88)	4.4° (6.70)	3.7° (5.36)	4.5° (6.54)	3.6° (5.42)	5.1° (7.77)
	Peak Extension Stance	5.0° (625)	4.4° (347)	4.2° (312)	4.5° (225)	5.6° (280)	5.4° (265)	4.5° (223)
	Peak Flexion Stance	4.7° (35.7)	5.0° (31.4)	7.1° (33.4)	6.7° (28.1)	7.3° (28.8)	8.0° (29.1)	4.7° (16.2)
Knee	Peak Flexion Swing	5.4° (9.47)	5.2° (8.81)	5.0° (7.89)	4.2° (6.49)	4.8° (7.41)	4.4° (6.99)	5.2° (8.22)
	Stance Excursion	3.8° (27.5)	4.9° (28.6)	4.9° (21.8)	6.5° (24.9)	4.4° (16.2)	4.9° (16.5)	4.1° (13.3)

Abbreviations: ICC (intraclass correlation coefficient); MDC (minimal detectable change); 2-D (two-dimensional); mph (miles per hour). Values within parentheses indicate MDC values as a percentage of mean scores.