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Accessing ankle kinematics in sagittal and frontal planes of motion by accelerometers during gait: an inexpensive alternative to optical motion analysis systems.

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Abstract

Background: An accelerometer-based system was developed to analyze the ankle kinematics in the sagittal and frontal planes of motion during gait. **Material and methods:** Two triaxial accelerometers were attached to the leg and foot of ten healthy volunteers. Gait kinematic data was compared with data obtained from an optical motion analysis system. **Results:** The pattern of the curves obtained using the accelerometer-based system is similar to that obtained by the optical motion analysis system, especially in the range of 0 to 60% of the gait cycle. Isolating the stance phase of gait, the coefficient of multiple correlation for the kinematic data of two systems was CMC = 0.99 for the sagittal plane and CMC = 0.97 for the frontal plane motion. The maximal RMS error in the stance phase of gait was 1.6 degrees in the sagittal plane of motion. **Conclusion:** The accelerometer-based system has low cost and accurate data of ankle kinematic in the stance phase of gait when compared to optical motion analysis systems.

Palavras-chave: Biomechanics, Gait kinematics, Accelerometer, Portable gait analysis, Optical motion analysis systems

Introduction

Several systems are available to quantify the movements of the lower limbs during human gait. Among the most used systems, can be cited the video-based and the sensors-based systems (Dugan, Bhat, 2005).

Equipment based on the triangulation of videos allows tracking and reconstruction of body segments in three dimensions for kinematics analysis. These systems have greater accuracy and higher cost which limits the clinical use, being habitually used in research (Mayagoitia *et al.*, 2002; Dugan, Bhat, 2005).

Systems based on sensors such as accelerometers and gyroscopes are portable, less complex, have until ten times lower cost, and provide similar quantitative data important to clinical assessment (Mayagoitia *et al.*, 2002; Aminian, Najafi, 2004; Wong *et al.*, 2007).

Currently, many studies concern the application of accelerometers to acquire kinematic data of lower limbs during gait. Those studies explore mainly the sagittal plane of motion, modulated gait speed situations, or sensors attached to articulated rigid structures, which are then fixed to the body segments [Williamson, Andrews, 2001; Mayagoitia *et al.*, 2002; Djurić-Jovičić *et al.*, 2011).

The purpose of this study was to develop a low-cost and simple system to analyze the motion of the ankle joint, based on accelerometers attached directly to the body segments of the volunteers, which is able to acquire angular data in the sagittal and frontal planes of motion, on self-selected gait speed.

Materials and Methods

Developed measurement system

For acquisition of the angular kinematics of the ankle during gait were used two triaxial accelerometers (Fig.1) (Model MMA7260Q, Freescale Semicondutor, Inc., Arizona, USA) attached to the shank and the foot of the volunteers by a belt.

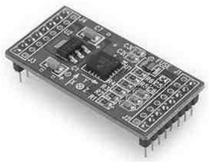

Figure 1 - Triaxial accelerometer Model MMA7260Q.

Those accelerometers provide a sensitivity of 800 mV/g, when supplied with 3.3V, and low values of acceleration, ±1g, when tilted from -90 to 90 degrees (Clifford, Gomez, 2005). The tilt angle (θ) was calculated according to Eq. (1):

$$\theta = \arcsin\left(\frac{V_{out} - V_{ref}}{800\,mV/g}\right) \quad (1)$$

where Vout is the accelerometer output in Volts and Vref is the accelerometer offset reference in Volts (Clifford, Gomez, 2005).

The relative angle was obtained by the difference between the absolute angles to each axis of motion.

Experimental Instrumentation

The A/D conversion was performed by a data acquisition board (DAQ USB-6009, National Instruments, Texas, USA) with 10 bits of resolution, in differential mode at a sampling rate of 200 Hz.

LabVIEW 2011 (National Instruments, Texas, USA) was used for the acquisition and storage of data from accelerometers activated by an external trigger (Fig.2).

Figure 2 - Acquisition apparatus for the accelerometers: **A.** Software developed in LabVIEW – **B.** Data acquisition board DAQ USB6009 – **C.** External trigger.

An optical motion analysis system composed of 10 ProReflex cameras (Qualisys Medical AB, Gothenburg, Sweden) was used to validate the data collected by the accelerometers.

Experimental Protocol

The study was approved by the ethical committee of the Universidade Federal de Minas Gerais (COEP-UFMG n° 362/09). The sample was composed of ten male volunteers, with no history of injuries in the ankle joint, with a mean age of 27.8 years and a BMI of 23.3. The volunteers signed a consent form prior to being included in the research.

Initially, anatomical passive markers were placed on the following points of the right lower limb: medial and lateral condyles of the femur, medial and lateral malleolus, calcaneus, and the head of the first and fifth metatarsals. A cluster containing three markers was fixed in the shank laterally and used for tracking this segment. The three anatomical passive markers that were attached to the calcaneus and metatarsals were used to track the foot. An accelerometer was attached to the side of the right shank of the volunteers with their axes X and Y parallels to the ground. A second accelerometer was attached to the dorsum of the volunteer's right foot supported by a wedge made from high-density ethylene vinyl acetate (E.V.A.) to ensure the alignment of its axes parallels to the ground (Fig.3).

Figure 3 – Positioning of passive markers and accelerometers on the volunteer.

The calibration of the accelerometers was performed before the beginning of each data acquisition. After the sensors were placed on the body segments and the volunteer assumed the orthostatic position, the voltage values in the accelerometers outputs (Vout) were recorded for 5 seconds. The voltage mean was used as a reference for the initial position in relation to gravity (Vref) for each sensor.

The two systems were synchronized by an external trigger (Fig.2C) and volunteers walked 10 times at a five-meter length walkway at a self-selected speed. Just the gait cycle executed in the central portion of the walkway was used for the analysis.

Processing and data analysis

The data acquired by the optical motion analysis system were processed in Visual 3D (C-Motion, Inc., Rockville, USA) to calculate the ankle kinematics in sagittal and frontal planes during gait, adopting the Visual3d 6 degrees of freedom model.

MatLab 2009 (MathWorks, Massachusetts, USA) was used to normalize and compare data from the two systems.

A 6th order Butterworth low-pass filter with a cut-off frequency of 5 Hz was applied to attenuate the high-frequency noise and obtain a smooth signal from accelerometers.

The angular values in both planes of motion obtained by the two systems were normalized in 101 points and were calculated as the averages and standard deviations of 100 gait cycles of volunteers.

In order to compare the curves generated by the systems, the coefficient of multiple correlation - CMC (Eq. (2)) calculated for the full gait cycle and for the range between 0 to 60% of the gait cycle was used.

$$\mathrm{CMC} = \sqrt{1 - \frac{\sum\sum(Y_{\mathrm{jt}}-\overline{Y_t})^2/T(N-1)}{\sum\sum(Y_{\mathrm{jt}}-\overline{Y})^2/(\mathrm{NT}-1)}}, \qquad (2)$$

where “T” is the number of samples and “N” is 2, the number of compared signals (Kadaba *et al.*, 1989).

To compare the average error of the measurements obtained by the system built with accelerometers in relation to the optical motion analysis system the root of the mean of the squared differences (Eq. (3)) calculated for the full gait cycle and for the range between 0 to 60% of the gait cycle was used.

$$RMS = \sqrt{\frac{1}{T}\sum_{i=1}^{T}\left(\varphi_{kr}(i) - \varphi_{k}(i)\right)^2} \qquad (3)$$

where subscript “r” indicates a reference measure and “” is the angle measurement (Mayagoitia *et al.*, 2002).

A percent error was calculated as the ratio between the RMS error to the average peak-to-peak amplitude of the optical motion analysis system (Mayagoitia *et al.*, 2002) calculated for the full gait cycle and for the range between 0 to 60% of the gait cycle.

Results

Fig. 4 presents the mean curve of the angular kinematics of the ankle in the frontal and sagittal planes during the gait cycle obtained by the optical motion analysis system and the system built with accelerometers.

The coefficient of multiple correlation calculated for the angular kinematics of the ankle in the sagittal plane was CMC = 0.82 and for the frontal plane was CMC = 0.97. When isolated the stance phase of gait, considered as the range between 0 to 60% of the gait cycle, the coefficient was CMC = 0.99 for the sagittal plane and CMC = 0.97 for the frontal plane of motion of the ankle.

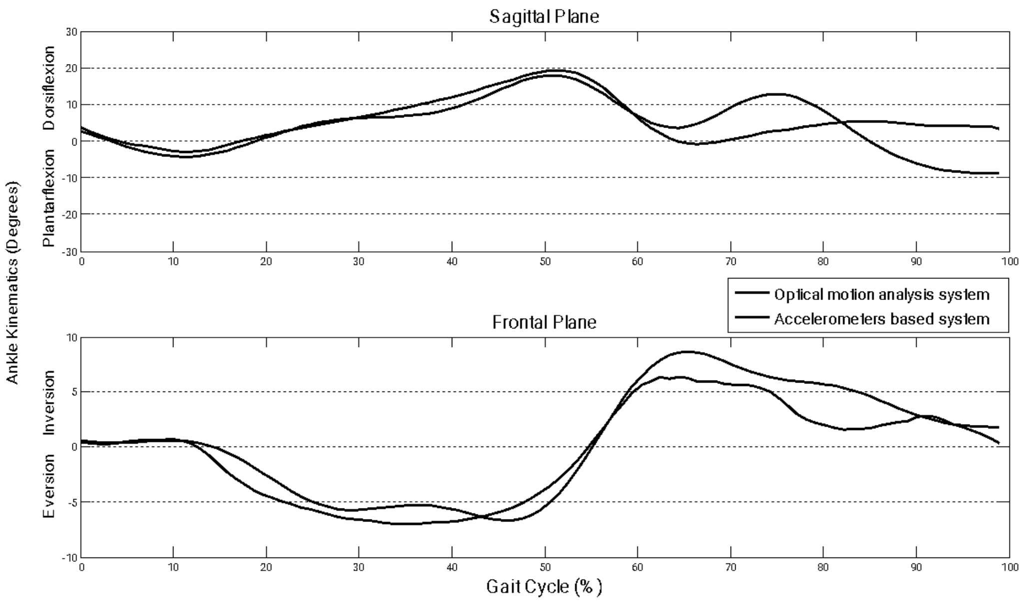

Figure 4 - Ankle kinematics of sagittal and frontal plane of motion in the gait cycle

The root of the mean of the squared differences, the percent error, and the signal average peak-to-peak amplitude can be seen in Table 1.

Table 1 - RMS error, peak-to-peak signal amplitude, and percent errors of the kinematic measurements.

	Sagittal Plane of Motion		Frontal Plane of Motion	
	Total gait cycle (0 to 100%)	Isolate gait cycle (0 to 60%)	Total gait cycle (0 to 100%)	Isolate gait cycle (0 to 60%)
RMS error (Degrees)	5.5192	1.5535	1.5890	1.0916
Signal amplitude (Degrees)	22.31	22.31	13.37	11.87
% Error	24.7	6.9	11.9	9.2

Discussion

As can be observed in Fig. 4, the pattern of the curves obtained by the accelerometer-based system is similar to that obtained by the optical motion analysis system in the range between 0 to 60% of the gait cycle for the frontal and sagittal planes of motion. This finding corroborates with the study conducted by Mayagoitia *et al.* (2002), however, in this work, was possible to evaluate two planes of motion consecutively and it was used a small number of sensors. Quantitatively, the accuracy of the system can be seen in Table 1, where the percent error for both planes of motion decreases on the stance phase of the gait isolated.

The interval between 61 to 100% of the gait cycle showed irregularities in the data, especially in the sagittal plane, when the two systems were compared. This phenomenon can be explained by the presence of high angular acceleration of the lower limb during the swing phase of gait (Williamson, Andrews, 2001). As the accelerometer is extremely sensitive to moments of sudden acceleration, in this phase of gait the signal obtained by the sensors loses accuracy (Clifford, Gomez, 2005).

Conclusion

Measuring systems built with accelerometers have low cost and wide applicability in the study of gait kinematics, however, to acquire more accurate data, further studies are needed to test the use of a third reference accelerometer for rejection mode analysis during the swing phase of gait and the application of another type of sensors like magnetic compass or gyroscopes to access the transverse plane of motion.

Conflict of interest: Authors state no conflict of interest.

Disclosure statement: No author has any financial interest or received any financial benefit from this research.

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