Acta Scientiarum

http://www.uem.br/acta
ISSN printed: 1806-2563
ISSN on-line: 1807-8664
Doi: 10.4025/actascitechnol.v36i1.18581

Design of a low cost measurement system based on accelerometers for gait analysis

Fermín Martínez-Solís[1*], Sergio Vergara-Limon[2], Aurora Vargas-Treviño[2], José María Rodríguez-Lelis[3], Abraham Claudio-Sánchez[1] and Marciano Vargas-Treviño[4]

*[1]Electrónica de Potencia, Interior Internado Palmira, s/n, 62490, Col. Palmira, Cuernavaca Morelos, México. [2]Facultad de Ciencias de la Electrónica, Benemérita Universidad Autónoma de Puebla, Grupo de Robótica, Puebla, México. [3]Ingeniería Mecánica, Interior Internado Palmira, Cuernavaca Morelos, México. [4]Escuela de Ciencias, Universidad Autónoma Benito Juárez de Oaxaca, Oaxaca, México. *Author for correspondence. E-mail: fsolis@cenidet.edu.mx*

ABSTRACT. Current research reports on the development of a portable electronic system to assess the kinematics of the lower limb joints at the sagittal plane. The electronic device characteristics and the different communication protocols to transfer data are also reported. Research obtained the hip and knee angles to analyze the lower limb kinematics during multiple gait cycles. Results showed that the movement patterns, found in the analysis made on people from central Mexico, were cyclical and alternating. The knee described a one third curve in flexion-extension movements just before the start of the flexion-extension curve in the swing phase. Moreover data obtained showed a correlation of movement between hip and knee during walking.

Keywords: electronic system, joints, lower limb, gait, movement pattern.

Projeto de um sistema de medição de baixo custo baseado em acelerômetros para análise de marcha

RESUMO. Este trabalho descreve o desenvolvimento de um sistema electrônico portátil para avaliar a cinemática das articulações dos membros inferiores no plano sagital. Além disso, as características dos dispositivos eletrônicos e os diferentes protocolos de comunicação para transferência de dados são relatados. O objetivo deste trabalho é obter os ângulos de quadril e joelho para análise cinemática dos membros inferiores durante os múltiplos ciclos de marcha. Os resultados mostraram que os padrões de movimento são cíclicos e alternados. Esses padrões foram encontrados na análise feita em pessoas do México central: o joelho descreve um terço de curva em movimentos de flexão-extensão pouco antes de iniciar a curva de flexão-extensão na fase de balanço. Além disso, os dados obtidos mostraram uma correlação de movimento entre quadril e joelho durante a caminhada.

Palavras-chave: sistema eletrônico, juntas, membros inferiores, marcha, padrão de movimento.

Introduction

Movement analysis has been the object of numerous studies that register some movements to estimate the kinematics in the lower limbs during gait. Pattern gait is quite variable; this variability is related to velocity, cadence, stride length and asymmetry patterns, even in a group with the same phenotype (VON-SCHROEDER et al., 1995). These studies analyzed locomotion to set down a pattern for healthy people. Since the particularities of gait pattern during the movement analysis may be useful to assess normal or pathological gaits, motion analysis studies are widely used in rehabilitation and in orthopedic and biomedical device design. In particularity, gait has been largely used in prosthesis and orthotics to improve the level in coordination between biomedical devices and the human body (HERR; WILKENFELD, 2003; POPOVIC; KALANIVIC, 1993).

Nowadays, there are many sophisticated biomechanical laboratories equipped with video cameras, force platforms and electromyography (EMG) systems. In these laboratories kinematics of the lower limbs may be obtained in two (2D) and three dimensions (3D), and may estimate the floor reaction forces and record the muscular activity during gait (MEDVED, 2001; PÉREZ et al., 1998; VILLA et al., 2008; WINTER, 2005). However, these laboratories are not only expensive and require complex instrumentation, but have only one environmental setting which only allows measurement within a restricted volume (CHE-

CHANG; YEH-LIANG, 2010; CORREA; BALBINOT, 2011; DEJNABADI et al., 2006; MAYAGOITIA et al., 2002; POLIDÓRIO et al., 1998).

New measurement systems for gait analysis have been developed in the last two decades. Developed systems with inertial sensors obtain 2D lower limb kinematics during several steps (CORREA; BALBINOT, 2011; DEJNABADI et al., 2005; KUN et al., 2011; MAYAGOITIA et al., 2002; PONS et al., 2007; WANG et al., 2010; WILLEMSEN et al., 1991). The systems based on accelerometers obtain the kinematics of lower limbs along a reference framework through acceleration in the body segments; the accelerometers fixed on body segments reflect the intensity and frequency in the movements (CHE-CHANG; YEH-LIANG, 2010). These systems have some advantages over biomechanics laboratories such as portability, low cost and the number of steps recorded. However, the biggest disadvantage has been getting the joint angles by acceleration. This is due to errors as offsets and angle drift generated by integrating the angular acceleration and angular velocity to obtain the angles of the hip and knee. Willemsen et al. (1991) concluded that the use of accelerometers have three potential errors caused by the reference system, accelerometers (error in measuring acceleration) and the proposed models. The last item is the most important.

In order to solve the problem associated with accelerometers, Mayagoitia et al. (2002) developed a system with accelerometers and gyroscopes to get the leg's angle and angular velocity. The method provided kinematic parameters without integrating the signal acceleration. The authors concluded that the method to get the kinematics using accelerometers and gyroscopes might be exact when compared to the camera system. In addition, Dejnabadi et al. (2005) reported a new approach to estimate the kinematics of the lower limbs in 2D via accelerometers without integrating the acceleration signal. Through virtual and physical sensors the joint angles were estimated at the sagittal plane which showed an RMS error of 1.30° with respect to a reference system (camera system). Kun et al. (2011) developed a measurement system with accelerometers and magnetometers to estimate the knee kinematics. The proposed method is based on two algorithms based 1) on measurement differences of fixed sensors; 2) on measurement differences of virtual sensors. Both were used to estimate the angles of flexion-extension, abduction-adduction and inversion-eversion.

Furthermore, GODWIN et al. (2009) evaluated the accuracy of the micro-electro-mechanical systems (IMS) by means of a pendulum. The analysis was performed in different situations: static, quasi-static and dynamic. The error for static and quasi-static conditions was 0.3° and for dynamic conditions between 1.9 and 3.5°. The IMS are accelerometer- and gyroscope-equipped systems.

The systems designed with inertial sensors try to obtain the lower extremity kinematics during gait. However, they have not yet reported the hip and knee angles of the two lower extremities during walking. Research works on movement analysis were developed to solve the problem of error caused by integrating the acceleration signal (CORREA; BALBINOT, 2011; DEJNABADI et al., 2005, 2006; KUN et al., 2011; MAYAGOITIA et al., 2002; PONS et al., 2007; WANG et al., 2010; WILLEMSEN et al., 1991).

Current research presents a portable low cost system based on accelerometers to capture the lower extremity joint motions. A method for obtaining the hip and knee angles was also reported. The method estimated the tilt angle of body segments using accelerometers placed on the skin surface of the lower limbs. The theoretical principle uses the gravity vector as reference point. Consequently, the tilt angle was obtained without integrating the acceleration through the acceleration polygon method. The measurement system records the hip and knee angles, without using filters, during multiple gait cycles over a mechanical treadmill. The purpose is to analyze the kinematics of the lower limbs. Results obtained made possible the setting of movement patterns and typical movements of correlation between joints, such as hip-hip and hip-knee, generated in gait.

Material and methods

Gait is a process of rhythmical movements of the lower limbs to move the center-mass toward forehead. These movements are carried out at a frequency ranging between 5 and 40 Hz (PONS et al., 2007).

A technique to analyze the movements of the lower extremity joints during gait uses inertial sensors (MEMS) such as accelerometers which are placed in the body segment focused for measuring the acceleration along an orthogonal system to obtain the tilt angle with respect to a reference point.

The operation principle of an accelerometer considers one test-mass suspended by springs, as shown in Figure 1 (KEMPE, 2011). The acceleration of the test-mass is the sum of the specific force and the force of gravity by unit mass. If m=1, the acceleration of the proof mass is:

$$\overrightarrow{a_m} = \overrightarrow{F_s} + g \tag{1}$$

Thus, the specific force is given by:

$$\overrightarrow{F_s} = k * \overrightarrow{d_y} = \overrightarrow{a_m} - \vec{g} \tag{2}$$

where:

k is the spring stiffness constant; $\overrightarrow{d_y}$ is the displacement caused by the stretching of the spring; $\overrightarrow{a_m}$ is the acceleration of the test-mass and $\vec{g}$ the gravity.

In Figure 1, the system's test-mass lies in static equilibrium; acceleration is zero. According to equation 2, the specific force is equal to gravity.

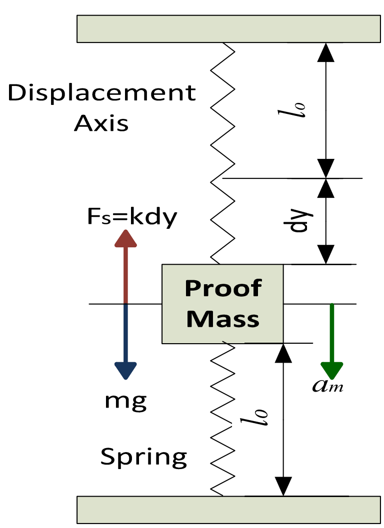

Figure 1. Operation principle of an accelerometer.

Now consider a two-axis accelerometer based on the design reported in (CHIH-MING et al., 2010) composed of a proof mass with two support frames and two groups of springs (Figure 2). The accelerometer is fixed to inertial frame X-Y where the acceleration is given by:

$$\vec{a} = \overrightarrow{a_m} + \vec{g} \tag{3}$$

where:

$\overrightarrow{a_m}$ is the acceleration of the test-mass of the accelerometer and $\vec{g}$ the gravity.

From Equation 2, the specified force to the accelerometer in Figure 2 is given by:

$$\overrightarrow{F_s} = k * \overrightarrow{d_y} = \vec{g}$$

In this case, the acceleration registered by the accelerometer is:

$$\vec{a} = \vec{g}$$

The accelerometer shown in Figure 2 is not perturbed by translation movements or relative rotation and the accelerometer measures the acceleration of gravity along the Y axis.

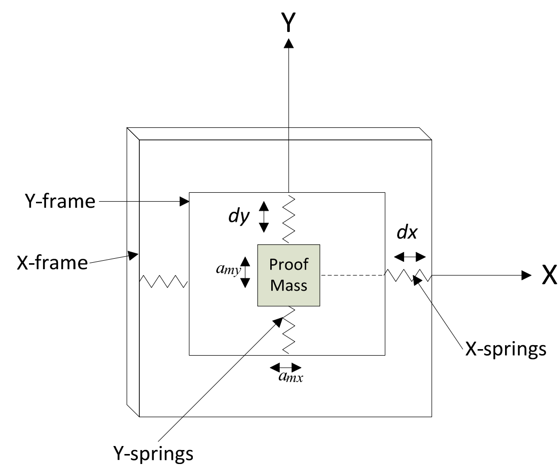

Figure 2. Physical model of a two-axis accelerometer.

As Equation 3 shows, the accelerometer measures the external acceleration (the acceleration of the body segment) by the acceleration of the test-mass (TITTERTON; WESTON, 2004).

To calculate the tilt angle in body segments (thigh and leg), a bar which represents a body segment on a new frame X-Y (where the joint center is the origin of the frame X-Y) is considered. An accelerometer is placed with frame x'-y' to calculate the tilt angle in the bar, as Figure 3 shows. The bar with accelerometer moves from point P1 to point P2 at time Δt (sample time); in this change, the accelerometer's frame (x'-y') suffers a displacement and a relative rotation with respect to initial frame (X-Y). The accelerations registered by the accelerometer for P1 and P2 are equal (see Equation 3). However, acceleration a_x changed at the two points:

1) P1, $a_x = 0$, therefore $\vec{a} = a_y = \vec{g}$; at this point the accelerometer's frame x´-y' is parallel to the initial frame X-Y, and the acceleration $\vec{a}$ is on the Y axis.

2) P2, $a_x \neq 0$; in this point if $|\vec{a}| = |\vec{g}|$, then $\vec{a}$ and

$\vec{g}$ have the same address, so $a_x \neq 0$ indicates a rotation with angle (θ) in the accelerometer frame x'-y', related to the initial frame X-Y.

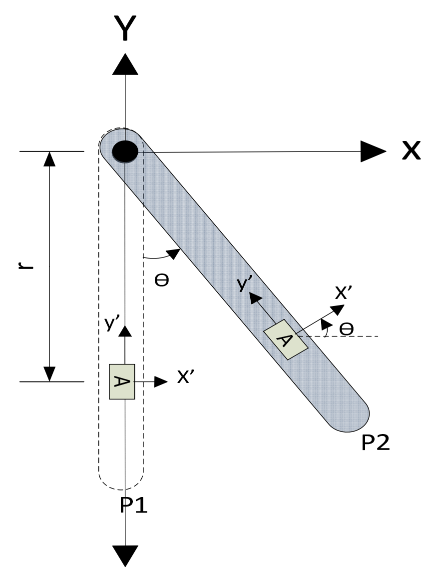

Figure 3. Body segment for measuring the tilt angle.

Thus the rotation angle θ of the accelerometer frame x'-y' at point P2 with respect to initial frame (X-Y, point P1) may be determined with acceleration polygon method, as shown in Figure 4.

So, the rotation angle θ is given by:

$$\theta = \tan^{-1}\frac{a_x}{a_y} \tag{4}$$

where:

a_x and a_y are accelerations measured by the accelerometer in frame x'-y'.

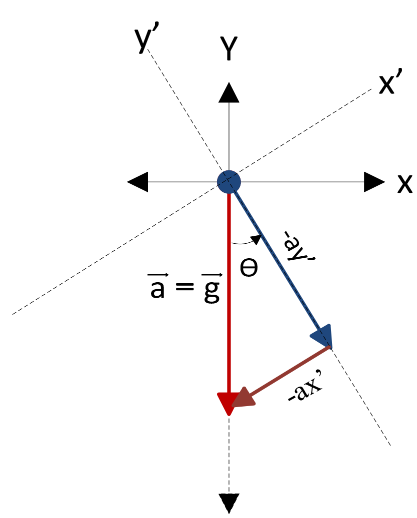

Figure 4. Acceleration Polygon Method

Based on the above, if $|\vec{a}| = |\vec{g}|$, the gravity vector may be used as reference point to measure the tilt angle. If $|\vec{a}| \rangle |\vec{g}|$, the two vectors do not have the same direction and the method is invalid in this case.

For lower limb kinematics, a new X-Y reference frame was established at the hip joint. The body segments were represented with links L1 and L2, as shown in Figure 5.

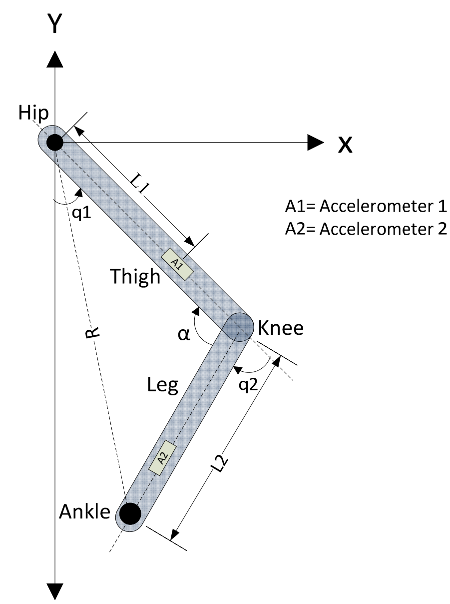

Figure 5. Estimate of angles for hip and knee.

According to Whittle (2007), the hip angle may be measured in two different ways: 1) the angle between the vertical and the femur which is known as the absolute angle, and 2) the angle between the pelvis and the femur, known as the relative angle. In current research, the angle hip was obtained by the first method: during a normal gait, the trunk stayed in a perpendicular position. In Figure 5, the angle $\theta 1$ is the tilt angle of the thigh and may be considered the absolute angle of the hip, given by:

$$q1 = \theta 1 \tag{5}$$

Similarly, the second method by Whittle (2007) was used to obtain the knee angle. First, the tilt angle of the leg (θ2) was obtained by accelerometer 2 (see Figure 5); from L1 projection, the relative angle of knee may be estimated by:

$$q2 = \theta 1 + \theta 2 \tag{6}$$

where:

$q2$ is a relative angle of the knee; θ1 is the tilt angle of the thigh; θ2 is the tilt angle of the leg.

Thus, the lower limb length may be determined by:

$$\alpha = \cos^{-1}\left(\frac{\|L1\|^2 + \|L2\|^2 - \|R\|^2}{2\|L1\|\|L2\|}\right) \tag{7}$$

$$R^2 = L1^2 + L2^2 \tag{8}$$

These determine the variation of the lower limb length for different angles of the hip and knee.

A reference accelerometer was not used in current research because the analysis was performed on a mechanical treadmill, i.e. the linear acceleration that experiences the body mass-center was not taken into account.

Measurement system

When accelerations and angles estimates are provided, the next step is to choose the devices to design the measurement system. Winter (2005) reported that the gait analysis developed over 71 Hz provides useful particularities to determine gait pattern. Therefore, the aim is to develop a measurement system with a sample rate above 71 Hz.

As Figure 6 shows, the block diagram of the measurement system has 1) four accelerometers MEMS as angle sensors; 2) PCI card for data acquisition (LEÓN-BONILLA et al. 2006); 3) a computer for data processing.

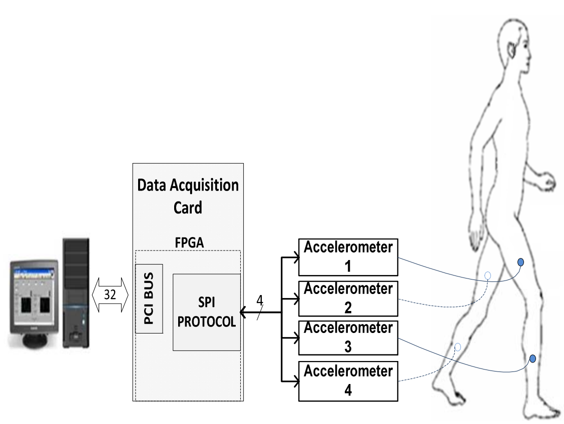

Figure 6. Block diagram of measurement system.

One accelerometer was placed at the lower extremity of each body segment to obtain the tilt angle (see Figure 3). A data acquisition card was also used to develop data transfer protocols and read the accelerometers, coupled to a PC to transfer and save data.

Some characteristics of the electronic devices used to develop the measurement system follow:

Sensors: Acceleration at plane X-Y was provided by four accelerometers (MEMS) with a band width of 1.5 KHz, a measurement range of ±1.7 gravities and 12 bits of resolution (the bit LSB is equal to 0.97 mg). Each accelerometer provided the acceleration via SPI protocol (Serial Peripheral Interface).

A configuration master-slave in the SPI protocol was developed to read the four accelerometers. In this case, the FPGA is the master device which generates the signals: 1) Clock SCK to synchronize the systems; 2) SDO to configure the slave devices; and 3) CS to select the slave devices. The slave devices send the acceleration signal by signal SDI.

PCI card: acquisition, processing and data flow control between PC and accelerometers were performed by PCI card with different data transfer protocols. PCI card is a modular card of open architecture with FPGA of 8256 logics gate which may work up to 100 MHz. Configuration and reading of each accelerometer and data transfer between accelerometers and PC was performed by an FPGA via firmware in different blocks.

1) PCI Block develops PCI protocol (to read-write) for data transfer of 32 bits between the acquisition card and the PC.

2) SPI Block develops the serial handshaking master-slave to communicate accelerometers and acquisition card.

3) Control Block synchronizes the different work frequencies of the blocks involved to keep the whole data during the information transference between PC and accelerometers.

Data processing: Software of virtual instrumentation in the PC was used to process, visualize and analyze the obtained data. The software allowed data transference by PCI bus in real-time with a rate sample of 250 Hz.

Figure 7 shows data transfer between PC and PCI card. The following tasks are developed for transfer:

a) Data acquisition and storage of the four accelerometers into FPGA: 32-bit packs are sent to FPGA via PCI bus to select, configure and read each accelerometer. The accelerometer then sends acceleration to FPGA by SPI protocol. This procedure is executed for the four accelerometers; consequently, FPGA temporarily saves the data of the four accelerometers for each sample.

b) Data transference of PC to FPGA: through software, a reading command is sent by the PCI bus to FPGA; in its turn, FPGA sends packs of 32 bits: the 16 most significant bits contain the label that indicates axis and the number of accelerometer, and the 16 less significant bits contain the acceleration of each accelerometer (see Figure 8).

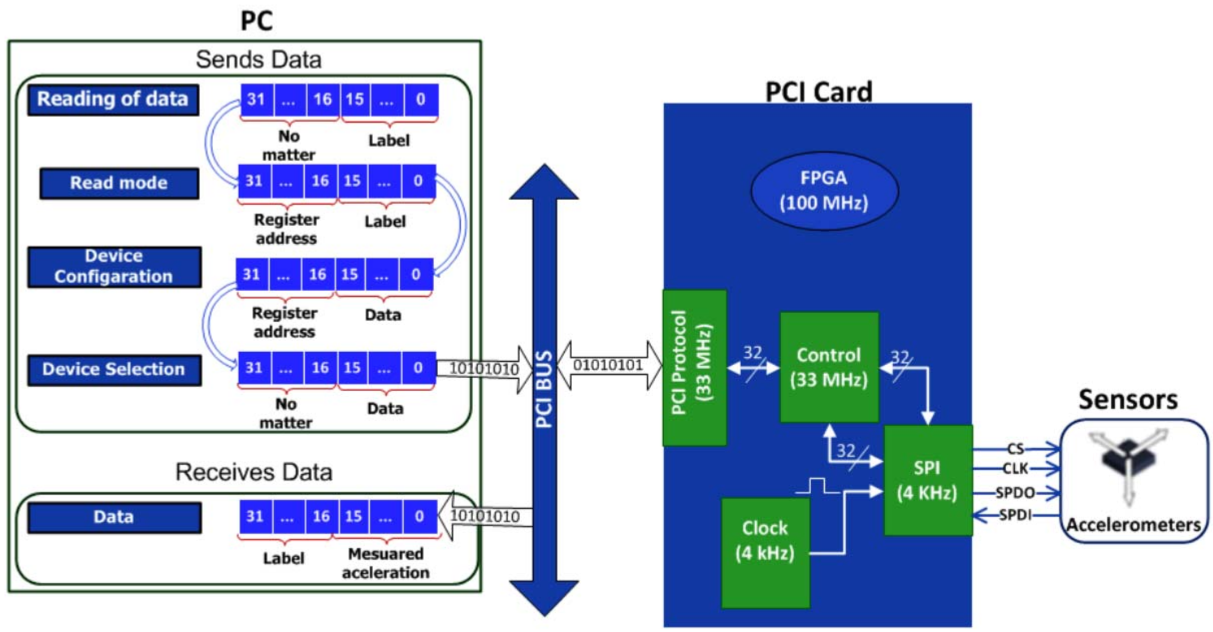

Figure 7. Data transfer for the reading of sensors.

c) The analysis and recording data: first, the data are shown on the PC to review the functioning of the measurement system; data are then recorded in a text file for analysis.

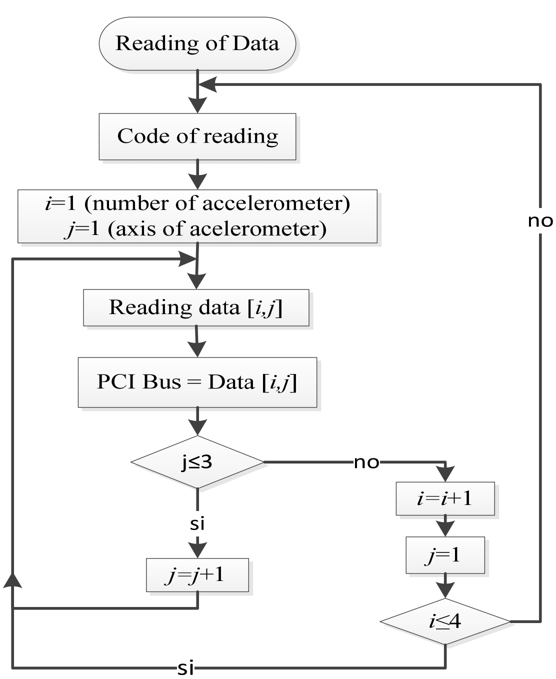

Figure 8. Reading PC-FPGA data.

Assessment of the measurement system

The method to obtain the tilt angle was assessed with an elbow of a manipulator robot of three degrees freedom reported by (REYES; ROSADO, 2004). A trajectory control was used only to control the manipulator robot's elbow movements, i.e. base and shoulder were fixed at zero degrees (without movement). The robot used an incremental encoder with 655,360 p/rev. as angle sensor (REYES; ROSADO, 2004). Figure 3 shows the accelerometer attached to the robot and its trajectory was determined by:

$$x = 90 * \sin(t) \tag{9}$$

An oscillation between 90 and -90 degrees (vertical is the reference or initial point) was developed by the elbow link of the manipulator robot with constant velocity of 1 rad/sec. The robot presented a position error of 0.68 degrees during the trajectory tracking.

Figure 9 presents accelerations in a_x and a_y registered by the accelerometer for movement of Equation 9; Y axis presents the acceleration ($m\ s^{-2}$), and X axis presents the time (second).

The continuous line represents a behavior of a_x acceleration, which starts at zero, and subsequently has an oscillation between -9.8 and 9.8 ($m\ s^{-2}$). Similarly, the dotted line represents behavior of a_y acceleration with an oscillation between -9.8 and 0 ($m\ s^{-2}$).

The tilt angle of the link robot was obtained by Equation 4 with a_x and a_y rates. Figure 10 shows the tilt angle which starts at zero, with a subsequent oscillation between 90 and -90 degrees.

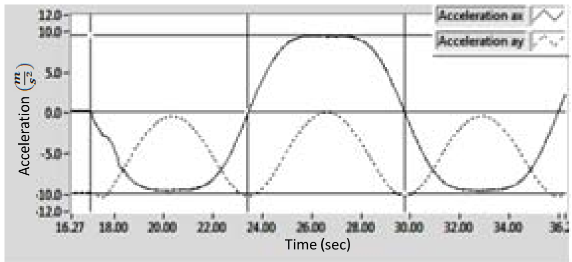

Figure 9. Accelerations: a_x continuous line; a_y dotted line.

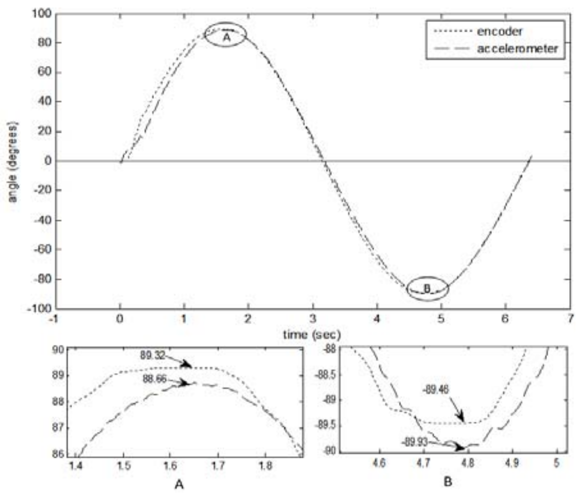

Figure 10. Elbow trajectory.

The tilt angle had an average error of 0.75 degrees when compared to the angle obtained by the motor encoder placed at the elbow joint of the manipulator robot.

Finally, Figure 11 presents the implementation of the electronic measurement system and the placement of accelerometers in the body segments.

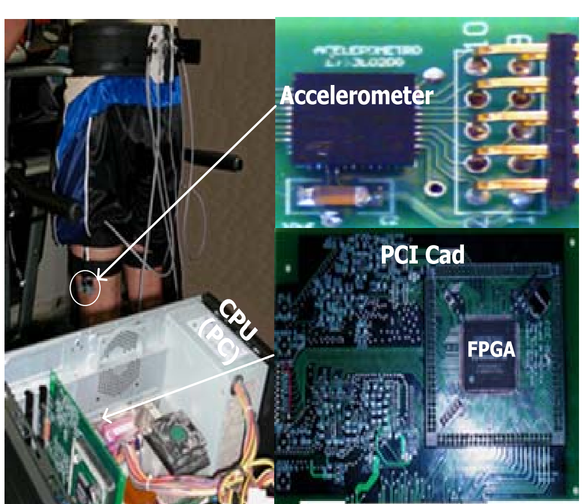

Figure 11. Measurement system.

Results

The gait analysis of eight healthy people, aged between 20 and 30 years, from different regions in Mexico, was evaluated on a passive treadmill with an average speed of 2 km h^{-1}. The length and weight of thigh and leg were obtained from anthropometric tables, taking into consideration the body's height and weight of each person (WINTER, 2005; SINGH et al., 2011). Data are shown in Table 1.

Table 1. Anatomic and anthropometric data of people analyzed.

	Body Weight (kg)	Body Height (m)	Thigh Length (m)	Leg Length (m)	Thigh Weight (kg)	Leg Weight (kg)
Person 1	64	1.75	0.43	0.40	6.3104	3.0016
Person 2	68	1.65	0.41	0.38	6.7048	3.1892
Person 3	78	1.67	0.41	0.39	7.6908	3.6582
Person 4	72	1.75	0.43	0.40	7.0992	3.3768
Person 5	64	1.64	0.40	0.37	6.3104	3.0016
Person 6	65	1.70	0.42	0.38	6.409	3.0485
Person 7	63	1.64	0.40	0.37	6.2118	2.9547
Person 8	59	1.63	0.40	0.37	5.8174	2.7671

The following were required for gait analysis: 1) four accelerometers: two on each lower limb, as in Figure 6; 2) a test for each person's adaptation to the measurement system; 3) standing motionless for 5 seconds at the beginning and at the end in each gait analysis; 4) recording the data in real-time for further analysis.

Figure 12 presents the hip angles and shows cyclic and alternating movements in both lower limbs. The alternation in the movements aims to keep the human body up and in equilibrium. Figure 12 also demonstrates that the movements of each lower limb are asymmetric.

The angles at the hip present a variation between 2 and 5 degrees at the maximum flexion angle in the gait cycles obtained during the test. This variation was associated with people's laterality, i.e. the dominant member presents a bigger flexion angle (see Figure 12). Furthermore, Figure 12 shows the gait cycle with two circles: 1) S.C indicates the initial contact in the stance phase and the start of the gait cycle, 2) E.C indicates the end in swing phase which shows the end of the gait cycle.

A dotted line rectangle represents the alternation (AL) in the lower limbs during gait. The alternation between lower limbs is presented in one half of each gait cycle when the left foot contacts the floor and the right hip registers a bigger extension (negative angle).

Moreover, Figure 13 exhibits the knee´s relative angles, the angles represent cyclic movements but with asymmetry in movement patterns. The latter failed to show any alternation such as that which occurred in the hip angles. In a gait cycle, the results showed two flexion movements with different amplitudes: one minor flexion in the stance phase and a greater flexion during the swing phase. Besides, in the knee angles, a particularity (in both knees) was registered, reported in Figure 13 with a dotted line semicircle. The particularity in both knees (P.R-K to right knee and P.L-K to left knee) had a different movement pattern at each gait cycle.

Figure 14 presents the natural motion of the correlation between hip and knee during gait. This correlation between the movements of flexion and extension in hip and knee may be: 1) C.S.P at the stance phase in the extension movements (semicircle of dotted line); 2) C.O.P in the swing phase at flexion-extension movements (semicircle of continuous line). In both cases, the knee follows the movements produced by the hip joint regardless of the manner. Figure 14 shows that there is no constant movement pattern in the gait phases.

This fact indicates that the brain continuously executes settings and tunings in the movement pattern of the knees, i.e. there are no two gait cycles with equal pattern movements.

Finally, Table 2 presents the parameters flexion and extension movement in hip and knee. Table 2 shows two new parameters: 1) the variation of maximum flexion, or rather, the difference in the angle of maximum flexion between right and left hip (this variation identifies people's laterality); 2) alternation time, or rather, the time between maximum flexion movements that indicate alternation in the lower limbs.

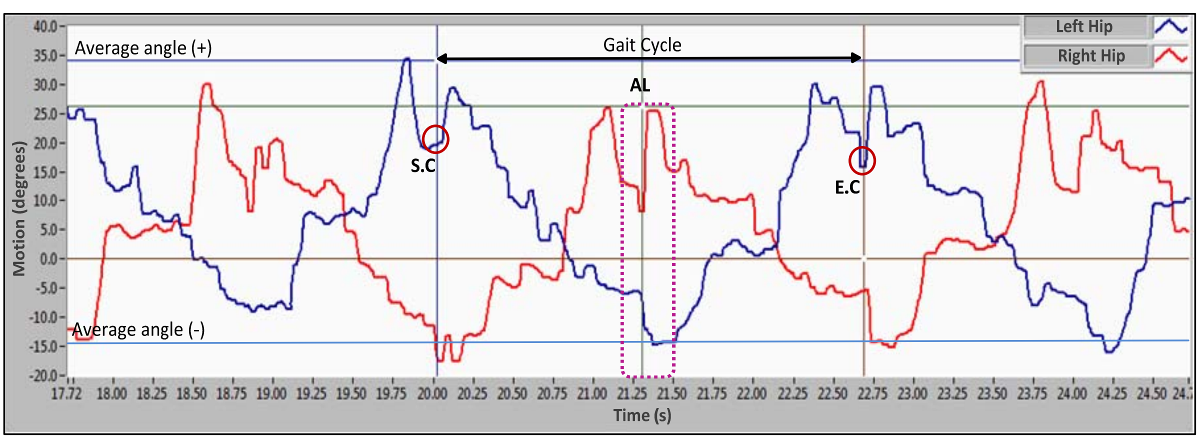

Figure 12. Hip angles (Person 1).

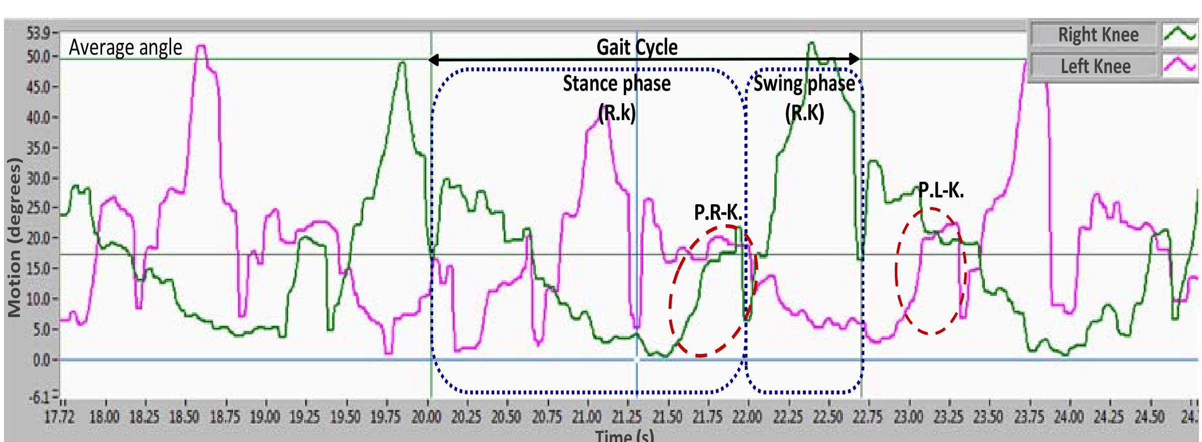

Figure 13. Knee angles (Person 1).

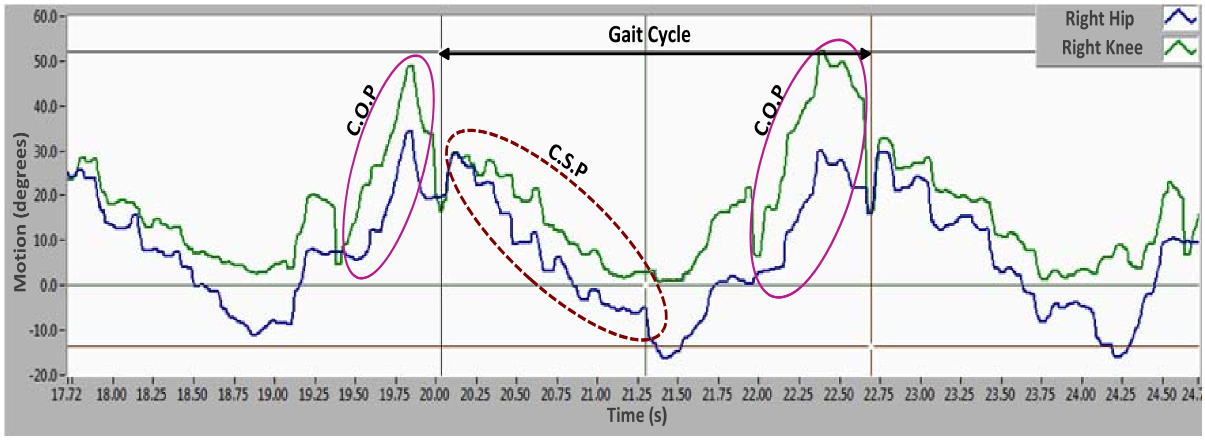

Figure 14. Movement patterns of hip and knee in the lower limb (Person 1).

Table 2. The parameters of people's gait under analysis.

	Person 1	Person 2	Person 3	Person 4	Person 5	Person 6	Person 7	Person 8
HIP								
Maximum flexion (degrees)	32	30	31	28	28.6	32	32	31
Maximum extension (degrees)	-13	-9	-9	-3	-4	-7	-5	-10
Variation of maximum flexion R L^{-1} (degrees)	6	2	4	5	3	2	6	3
Time of alternation R L^{-1} between maximum flexion (seconds)	1.4	1.2	1.2	1.1	1	1.2	12	1
KNEE								
Flexion 1 (degrees)	22	21	22	16	21	22	18	19
Extension 1 (degrees)	4	3	3	3	4	4	3	3
Flexion 2 (degrees)	42	43	43	43	44	44	40.1	48
Extension 2 (degrees)	4	2	4	4	3	3	4	3

Discussion

Measurement systems for gait analysis are currently expensive, especially the instrument-equipped systems with cameras. Moltenbrey (2009) published the average costs of one system with eight cameras and motion tracking software which is ideal for human movement analysis. The systems developed with inertial sensors have lower costs. One system with five inertial sensor modules for human movement analysis costs 57% less than the camera system. High costs of electronic measurement systems are the biggest problem for the development of new studies related with human movement analysis which are used widely in rehabilitation, orthopedic, biomechanics and design of prosthesis and orthotics for the lower limbs.

Current research presented one low cost measurement system of gait analysis. The system with four accelerometers, one PCI card and one PC Pentium IV with PCI slot costs only 7.6% of the camera system reported above. A method has also been reported to obtain the tilt angle from each body segment without integrating the acceleration signal. The angles obtained have some errors associated with three factors: 1) the alignment of the accelerometer with the vertical; 2) the small displacement of the accelerometer during analysis; 3) the error in the method to obtain the tilt angle. In fact, the maximum error in the hip angle was 1.75 and 3.5° in the knee angle, the latter being higher because the angle knee depended on the tilt angles of the thigh and the leg.

Research by Willemsen et al. (1991), Mayagoitia et al. (2002), Dejnabadi et al. (2005, 2006) and Kun et al. (2011) employed more than one accelerometer to calculate the angle of one joint (hip or knee). Furthermore, some of these studies have used magnetometers and virtual sensors to estimate the joint angles. The above papers only reported the evolution of the angle and the acceleration signal without identifying the gait cycle.

Djuric-Jovicic et al. (2011) reported a method to estimate joint angles throughout a system, developed with accelerometers, which do not need to integrate the acceleration to get the joint angles. They analyzed the gait of 27 healthy subjects on a treadmill. The error of the system was 6° evaluated by goniometers.

Current research estimated the kinematics of the lower limbs and identified the gait cycle and some of its features with the advantages and disadvantages of accelerometers. Movement patterns were obtained without using filters, similar to other works (DEJNABADI et al., 2005, 2006; DJURIC-JOVICIC et al. 2011; KUN et al., 2011;

MAYAGOITIA et al., 2002; WILLEMSEN et al., 1991).

On the other hand, several research works on gait analysis have reported a standard pattern in the movements of the joints of the lower limbs during gait. Donal (2002) reported movement patterns for hip and knee at the sagittal plane with the following characteristics:

1) with regard to the hip, there was an average amplitude in the flexion angle of 30° and an average extension angle of -10°.

2) with regard to the knee, there were two flexion movements: one in the stance phase with an average angle of 20° and another in the swing phase with an average angle of 60°.

Current research specifically reported the movement patterns on right and left hips which are similar to that reported by Donal (2002), as may be seen in Figure 12.

Further, Figure 12 shows small peaks (maximum angle variation in flexion) related to the walking style of each person i.e. each person has a particular form of ground contact. The small peaks reveal the end of the swing phase and the start of the stance phase. These particular movements in hip and knee reveal the start and end in each gait cycle. Moreover, the movement patterns demonstrated that the two gait cycles were different.

The movement patterns in the knee show two flexion movements on the sagittal plane (one in the stance phase and another in the swing phase), as reported by Donal (2002). However, the movement patterns do not have the same form or the particular movements that are shown in Figure 13. One disadvantage of a standard gait is the loss of the particularities that define the style of motion in a group of people, all of whom have a different phenotype.

The motion pattern in the knee during gait is related directly with the hip joint movement. In fact, it governs the amplitude and speed for each movement in the gait. In accordance with results in Figure 14, the hip acts as the joint that connects the lower limbs with the upper body, i.e. the lower limbs act as extensions that follow the hip´s movements during gait.

Conclusion

Current research presents a low cost portable system that allows the verification of movement patterns of the joints in lower limbs during the gait. The system demonstrated that the movements are cyclical and alternating, as reported in several studies of gait analysis developed in laboratories of biomechanics.

During a normal gait developed on a flat floor, the trunk maintains a perpendicular position. However, if for some reason it changes its position, the movements patterns of hip and knee change due to the position change in the mass center, i.e. the hip and knee angles may increase or decrease depending on the position of the body mass center. It is highly important to know the trunk position to determine whether the movements correspond to a normal gait over a flat floor or a gait with specific characteristics caused by the terrain or by the trunk's particular position. Variations in the hip and knee angles caused by the trunk position may be used to classify the normal gait, the gait on different terrains, and activities where the trunk has a particular position. These situations are beyond the scope of this work but they may be the object of future research work. Current paper does not consider the use of an accelerometer at the trunk. Consequently, the method to obtain the hip angle is valid only when the trunk is in a perpendicular position. According to the above, the hip angle was obtained with respect to verticality and it is considered an absolute angle. Since the knee angle was obtained between the femur and the tibia, a relative angle was obtained.

Results showed that the motion pattern during gait is associated with people's phenotype.

References

CHE-CHANG, Y.; YEH-LIANG, H. A review of accelerometry-based wearable motion detectors for physical activity monitoring. **Sensors**, v. 10, n. 8, p. 7772-7788, 2010.

CHIH-MING, S.; MING-HAN, T.; YU-CHIA, L.; WEILEUM, F. Implementation of a monolithic single proof-mass tri-axis accelerometer using CMOS-MEMS technique. **IEEE Transaction on Electronic Devices**, v. 57, n. 7, p. 1670-1679, 2010.

CORREA, D.; BALBINOT, A. Acceleromatry for the motion analysis of the lateral plane of the human body during gait. **Health Technology**, v. 1, n. 1, p. 35-46, 2011.

DEJNABADI, H.; JOLLES, B. M.; AMINIAN, K. A new approach to accurate measurement of uniaxial joint angles based on a combination of accelerometers and gyroscopes. **IEEE Transaction on Biomedical Engineering**, v. 52, n. 8, p. 1478-1484, 2005.

DEJNABADI, H.; JOLLES, B. M.; CASANOVA, E.; FAU, P.; AMINIAN, K. Estimation and visualization of sagittal kinematics of the lower limbs orientation using body-fixed sensors. **IEEE Transaction on Biomedical Engineering**, v. 53, n. 7, p. 1385-1393, 2006.

DJURIC-JOVICIC, M.; JOVICIC, N.; POPOVIC, D. Kinematics of gait: new method for angle estimation based on accelerometers. **Sensors**, v. 11, n. 11, p. 10571-1585, 2011.

DONAL, A. **Kinesiology of the Musculoskeletal System**. Orlando: Mosby, Inc., 2002.

GODWIN, A.; AGNEW, M.; STEVENSON, J. Accuracy of inertial motion sensors in static, quasistatic and complex dynamic motion. **Journal of Biomechanical Engineering**, v. 131, n. 11, p. 114501,1-114501,5, 2009.

HERR, H.; WILKENFELD, A. User-adaptative control of a magnetorheologicalprosthetic knee. **Industrial Robot: An International Journal**, v. 30, n. 1, p. 42-55, 2003.

KEMPE, V. **Inertial MEMS**: Principles and practice. Cambridge: Cambridge University Press, 2011.

KUN, L.; INOUE, Y.; SHIBATA, K.; ENGUO, C. Ambulatory estimation of knee-joint kinematics in anatomical coordinate system using accelerometers and magnetometers. **IEEE Transaction on Biomedical Engineering**, v. 58, n. 2, p. 435-442, 2011.

LEÓN-BONILLA, A.; VERGARA, S.; VARGAS, M. A.; REYES, F. Tarjeta PCI para control de un robot con tres grados de libertad. **CIINDET, IEEE Sección Morelos y la Asociación Mexicana de Ingenieros y Electricistas A. C**. México: CIINDET, IEEE, 2006.

MAYAGOITIA, R.; NENE, A.; VELTINK, P. Accelerometer and rate gyroscope measurement of kinematics: an inexpensive alternative to optical motion analysis systems. **Journal of Biomechanics**, v. 35, n. 4, p. 537-542, 2002.

MEDVED, V. **Measurement of human locomotion**. Boca Raton: CRC, PressLLC., 2001.

MOLTENBREY, K. Spotlight: your resource for products, user publications, news, and market research. **Computer Graphics World**, v. 32, n. 12, 2009. Available from: <http://www.cgw.com>. Access on: Feb. 4, 2013.

PÉREZ, J.; PICHARDO, A.; CHAVEZ, D. Análisis de parámetros cinemáticos de la marcha normal. estudio y análisis de patrones de marcha. **Revista Mexicana de Traumatología y Ortopedia**, v. 12, n. 5, p. 373-376, 1998.

POLIDÓRIO, A. M.; BORTOLOZZI, F.; FIGUEIREDO, M. Calibracao geométrica de cameras, aplicando o método dos mínimos quadrados. **Acta Scientiarum**, v. 20, n. 4, p. 495-503, 1998.

PONS, J. L.; MORENO, J. C.; BRUNETTI, F. J.; ROCON, F. J. Lower limb wearable exoskeleton. **Rehabilitation Robotics**, v. 4, n. 2, p. 471-498, 2007.

POPOVIC, D. B.; KALANOVIC, V. D. Output space tracking control for above-knee prosthesis. **IEEE Transaction on Biomedical Engineering**, v. 40, n. 6, p. 549-557, 1993.

REYES, F.; ROSADO, A. Polynomial family of PD-type controllers for robot manipulators.**Control Engineering Practice**, v. 13, n. 4, p. 441-450, 2004.

SINGH, S.; SINGH, K.; SINGH, M. Characteristics and body types of high performer and low performer hammer throwers. **Brazilian Journal of Biomotricity**, v. 5, n. 2, p. 80-86, 2011.

TITTERTON, D.; WESTON, J. L. **Strapdown Inertial Navigation Technology**. 2nd ed. Reston: The American Institute of Aeronautics and Astronautics, 2004.

VILLA, A.; GUTIÉRREZ, E.; PÉREZ, J. C. Consideraciones para el análisis de la marcha humana. Técnicas de videogrametría, electromiografía y dinamometría. **Revista de Ingeniería Biomédica**, v. 2, n. 3, p. 16-26, 2008.

VON-SCHROEDER, H.; COUTTS, R.; LYDEN, P.; BILLINGS, E.; NICKEL, V. Gait parameters following stroke: A practical assessment. **Journal of Rehabilitation Research and Development**, v. 32, n. 1, p. 25-31, 1995.

WANG, N.; REDMOND, S. J.; AMBIKAIRAJAH, E.; CELLER, B. G.; LOVELL, N. H. Can triaxial accelerometry accurately recognize inclined walking terrains? **IEEE Transaction on Biomedical Engineering**, v. 57, n. 10, p. 2506-2516, 2010.

WHITTLE, M. W. **Gait Analysis an Introduction**. 4th. Ed. Suzhou: Elsevier Ltd, 2007.

WILLEMSEN, A.; FRIGO, C.; BOOM, H. Lower extremity angle measurement with accelerometers-error and sensibility analysis. **IEEE Transaction on Biomedical Engineering**, v. 30, n. 12, p. 1186-1193, 1991.

WINTER, D. **Biomechanics and motor control of human movement**. Ontário: Wiley, 2005.

Received on September 11, 2012.
Accepted on March 19, 2013.