Kinematic Gait Analysis Using Inertial Sensors with Subjects after Stroke in Two Different Arteries

Article *in* Journal of Physical Therapy Science · August 2014

DOI: 10.1589/jpts.26.1307

9 authors, including:

Bruno Mazuquin
Manchester Metropolitan University

João P. Batista Jr
SRH University of Applied Health Sciences

Ligia M Pereira
Universidade Estadual de Londrina

Josilainne Dias
Universidade Estadual de Mato Grosso do Sul

J. Phys. Ther. Sci.
26: 1307–1311, 2014

Case Study

Kinematic Gait Analysis Using Inertial Sensors with Subjects after Stroke in Two Different Arteries

Bruno Fles Mazuquin[1)], João Pedro Batista Junior[1, 2)], Ligia Maxwell Pereira[1)], Josilainne Marcelino Dias[1)], Mariana Felipe Silva[1)], Rodrigo Luiz Carregaro[3)], Paulo Roberto Garcia Lucareli[4)], Felipe Arruda Moura[5)], Jefferson Rosa Cardoso[1)*]

[1)] *Laboratory of Biomechanics and Clinical Epidemiology, PAIFIT Research Group, Universidade Estadual de Londrina: Av. Robert Kock 60, 86035-440 Londrina, PR, Brazil*
[2)] *Instituto Federal do Paraná, Campus Londrina, Brazil*
[3)] *Universidade de Brasilia, Brazil*
[4)] *Department of Rehabilitation Science, Human Motion Analysis Laboratory, Universidade Nove de Julho, Brazil*
[5)] *Sports Science Department, Universidade Estadual de Londrina, Londrina, Brazil*

Abstract. [Purpose] The aim of the present study was described the kinematic characteristics of gait in stroke patients with two different arteries involved. [Subjects and Methods] Two patients who had suffered a basilar (A) or middle (B) cerebral artery ischemic stroke were compared with a control (C). Seventeen inertial sensors were used with acquisition rate of 120 Hz. The participants walked 3 times on a 10 meter walkway. From the raw data, the three gait cycles from the middle of each trial were chosen and analyzed. [Results] During the stance phase, patients A and B had a lower hip angle at initial contact and maximum flexion angle during load response than the control. Patient A and the control subject had similar knee angle values at initial contact, and patient B presented a flexed position in the initial phase of the gait cycle. The maximum flexion angles during loading response were also higher for patient B. The sagittal plane excursion for the ankle joint was lower for patient B in comparison with the other subjects. [Conclusion] Differences during walking between patients who had stroke in different arteries may be related to an alternative compensatory strategy. Patient A and the control subject had similar gait cycle curves at all joints, while patient B showed a rigid synergic pattern.

Key words: Stroke, Gait, Kinematic

(This article was submitted Jan. 9, 2014, and was accepted Feb. 7, 2014)

INTRODUCTION

The joint kinematics of patients with hemiparesis exhibit differences from the normal conditions in both the stance and swing phases of gait and also large interindividual variability[1)]. Healthy people are able to keep their bodies balanced on irregular surfaces, which can be explained by a neural strategy leading to dynamic stability[2)]. The postural adjustments that occur in the young and the elderly are specific to each task and may vary according to ground and environment factors. Postural balance, therefore, is relevant clinical evidence after neuromuscular impairments, such as in the case of stroke[3, 4)].

An important clinical finding is the kinematic compensatory pattern caused by neuromuscular impairment. Each patient's gait cycle varies according to the severity of the impairment and also their ability to readapt to new conditions and difficulties[5)]. The resultant gait pattern is a combination of the deviations caused by the primary dysfunction, compensatory movements, and the residual function. A common deviation is hip hiking or circumduction in the swing phase due to inadequate strength of the ankle dorsiflexors and subsequent difficulty of the foot adaptation seen in an elevation of the pelvis instead of hip flexion. Another common feature of the stroke gait is excessive knee flexion in the stance phase that is a consequence of diminished strength, which causes diminished support moment generation at appropriate joint angles by the combination of ankle plantar flexors and knee and hip extensors; thus, the knee may flex excessively[6)].

To ensure appropriate intervention strategies, professionals must be able to detect compensatory mechanisms. Thus, therapeutic procedures may focus on the primary cause and not the most known compensation. A new system based on inertial sensors is being used to assess gait kinematics, without cameras or bone markers, enabling subjects to walk freely indoors or outdoors. The sensors are composed of gyroscopes, accelerometers, and magnetometers, which are combined in order to obtain data such as acceleration, angular velocity, and joint angle[7)]. This system can discriminate gait symptoms, demonstrating a practical

*Corresponding author. Jefferson Rosa Cardoso (E-mail: jeffcar@uel.br)

approach to evaluation of human locomotion[8, 9]. Thus, the purpose of this case report was to analyze gaits of patients who had strokes with different etiologies and to compare kinematic data from them with those of a subject without impairments. A second purpose was to describe and observe the applicability of the new gait analysis system using inertial sensors.

SUBJECTS AND METHODS

In this case series study, two poststroke patients (A and B) and one healthy subject (C) were compared. Patients A and B had suffered a basilar and middle cerebral artery ischemic strokes, respectively, with right paresis. To characterize the motor impairment of the patients, the Barthel Index was used (100 represents the highest independence). The anthropometric characteristics, length of the injury, and Barthel Index of the volunteers are shown in Table 1. To participate in this study, the individuals needed to be male, between 40 and 65 years of age, and free from the need for any assistive devices and had to have been affected by only one event with a minimum post-injury time of three years[10]. This study was previously approved by the University Ethics Committee-UEL (#177/2011), and the volunteers gave written informed consent prior to participation.

The experimental procedures were conducted at the Laboratory of Biomechanics and Clinical Epidemiology. The inertial sensors system (MVN, Xsens Tech®, Enschede, Netherlands) was used to evaluate the kinematic data through a 3D analysis system of human gait movement. Seventeen inertial sensors (38 × 53 × 21 mm), 30 grams each, were attached to the whole body with Velcro® strips (except the head sensors) and connected with cables to an integrator system (Xbus Master, Xsens Tech®, Enschede, Netherlands) fixed to the back at the waist; the whole system weighs 1.9 kg. The Xbus Master communicated with two receptors connected to a personal computer using a Bluetooth system (RS-232/USB). Calibration was performed according to the specifications in the manufacturer's manual using anthropometric measures. Individuals were instructed to adopt an orthostatic and anatomic position for a few seconds, allowing software recognition of the body. Before the test began, the subjects were familiarized with all procedures and instructed to walk three times on a ten meter walkway at a self-selected speed in a predetermined direction.

The MVN Studio software (Xsens Tech®, Enschede, Netherlands) was used to collect (acquisition rate of 120 Hz), visualize, and save the data. From the raw data, the three gait cycles from the middle of each trial were chosen and analyzed, considering the initial contact of the foot until the initial contact of the impaired contralateral one. The best cycles were chosen, excluding the first and the last cycles, thus excluding also the positive and negative acceleration periods. The three selected gait cycles were imported into the MATLAB software (R2009a, MathWorks, Natick, MA, USA) and averaged as function of the cycle percentage. The Euler angles were used and calculated according to spatial position between each sensor. The average right hip, knee, and ankle joint angles of all subjects were plotted as gait cycle percentages in the sagittal plane. The joint angle values were collected and described in Table 2[11].

RESULTS

The results are described below according to the analysis of the joint curves plotted in the sagittal plane (Figs. 1–3). The angle values in specific moments of the gait cycle are shown in Table 3.

During the stance phase, patients A and B had lower hip angles at initial contact (H1) and maximum flexion angles during load response (H2) than the control (C). The maximum extension angles in the stance phase (H3) presented different peak values for both patients and this value for patient A occurred earlier than for the other subjects. The control subject also showed a higher angle value at toe-off (H4) than the other subjects, representing a normal extension pattern. Patient B had a similar maximum flexion angle (H5) during the swing phase in comparison with the control. The sagittal plane excursion (H6) of patient B was lower than those of the other subjects.

Patient A and the control subject had similar knee angle values at initial contact (K1), and patient B presented

Table 1. Subject characteristics

	Patient A	Patient B	Control
Age	57	63	56
Height (cm)	168	170	171
Mass (kg)	70	76	93
Duration of injury (years)	6	4	–
Barthel Index	100	85	–

Table 2. Variable descriptions

Joint	Variables
Hip	H1, hip angle at initial contact; H2, maximum flexion angle during loading response; H3, maximum extension angle in stance phase; H4, hip angle at toe-off; H5, maximum flexion angle in swing phase; and H6, total sagittal plane excursion.
Knee	K1, knee angle at initial contact; K2, maximum flexion during loading response; K3, maximum extension in stance phase; K4, knee angle at toe-off; K5, maximum flexion angle in swing phase; and K6, total sagittal plane excursion.
Ankle	A1, ankle angle at initial contact; A2, maximum plantar flexion angle during loading response; A3, maximum dorsiflexion in stance phase; A4, ankle angle at toe-off; A5, maximum dorsiflexion angle in swing phase; A6, total sagittal plane excursion, and A7, maximum plantar flexion angle in swing phase.

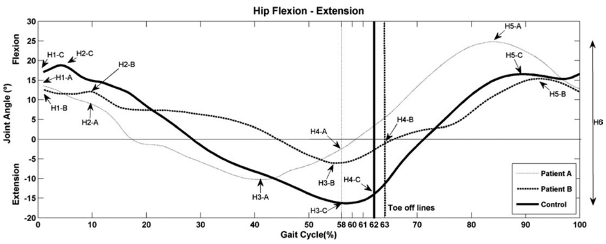

Fig. 1. Hip angles during the gait cycle.

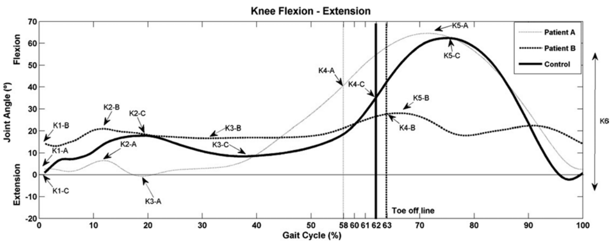

Fig. 2. Knee angles during the gait cycle.

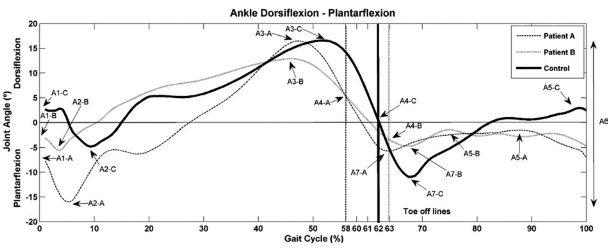

Fig. 3. Ankle angles during the gait cycle.

a flexed position in the initial phase of the gait cycle. The maximum flexion angles during loading response (K2) were also higher for patient B, and at maximum extension in the stance phase (K3), patient B kept a flexed knee position. At toe-off (K4), patient A and the control subject had similarities in angle values. During the swing phase, patient A and the control subject C presented similar values for maximum flexion (K5). The sagittal plane excursion (K6) was lower for patient B in comparison with the other subjects.

The stance phase started with different values (A1) for all subjects, with the highest difference between patient A and the control. The control subject presented the lowest maximum plantarflexion angle during loading response (A2) in comparison with patients A and B. The maximum dorsiflexion angle in the stance phase (A3) was similar for patient A and the control was the same for A4; patient B showed lower value in A3 and a higher value in A4. During the swing phase, patient B and subject C showed similar dorsiflexion peak angles, but patient A showed a greater peak angle. The sagittal plane excursion (A6) for the ankle joint was lower for patient B in comparison with patient A and the control. The highest plantarflexion angle value (A7) was presented by the control subject, while the patients presented different values.

DISCUSSION

The aims of this study were to describe and observe the applicability of kinematic gait analysis using inertial sensors for two stroke patients with different arteries compromised compared with a subject without impairments.

Table 3. Hip, knee, and ankle angles (°) in the sagittal plane

		Patient A	Patient B	Control
Hip Joint				
Hip angle at initial contact	H1	13.5	12.5	17
Maximum flexion angle during loading response	H2	8.9	12.3	18.6
Maximum extension angle in stance phase	H3	10.4	7.8	16.4
Hip angle at toe-off	H4	0.8	2.4	14.1
Maximum flexion angle in swing phase	H5	24.9	15.9	16.6
Total sagittal plane excursion	H6	35.4	23.8	33.1
Knee Joint				
Knee angle at initial contact	K1	2.55	14.2	1.6
Maximum flexion during loading response	K2	6.7	22.2	18
Maximum extension in stance phase	K3	0.9	16.3	8.2
Knee angle at toe-off	K4	38	29.9	36.8
Maximum flexion angle in swing phase	K5	64.9	32.2	62.7
Total sagittal plane excursion	K6	67.4	46.5	64.3
Ankle Joint				
Ankle angle at initial contact	A1	7.5	3.1	2.5
Maximum plantarflexion angle during loading response	A2	17	5.7	5
Maximum dorsiflexion in stance phase	A3	16.5	13	16.7
Ankle angle at toe-off	A4	1.7	4.9	1.9
Maximum dorsiflexion angle in swing phase	A5	2	0.6	0.9
Total sagittal plane excursion	A6	23	18.6	29.8
Maximum plantar flexion angle in swing phase	A7	6.4	5.5	13.1

Inertial sensors could be a valuable tool for objective gait analysis when disorders or impairments are present. There may be discrepancies between a patient's main complaint, such as a deficit in muscle strength, and the kinematic and kinetic gait patterns, demonstrating the relevance of a functional analysis, such as that in the present study[12)].

In this study, the patients who had suffered a stroke had different gait patterns, but patient A behaved similarly to the healthy subject (the control). A synergistic gait pattern is observed when patients with stroke are walking, unlike healthy individuals, who present selective motor control of joint movements[12)]. During walking, patients who have had a stroke normally show two synergic patterns: an extension mass pattern during the stance phase caused by synergic contraction of the quadriceps and gluteus maximus and a flexion mass pattern during the swing phase caused by synergic contraction of the hip, knee, and ankle flexors[13)].

These functional events can be noticed during case studies. Some other differences could be observed, like differences in walking velocity, stride length, and gait cycle durations, but the attempt was to observe maximal flexion and extension values and synergic patterns[14)]. The step length was found to be asymmetrical as a result of the shorter stance phase and longer swing phase in the paretic limb compared with in the healthy individual. In addition, it is important to point out that velocity was not controlled in this study, so the toe-off angle, considered a "marker" in the gait cycle, may indicate approximate values for the control subject.

At the initial contact, patients A and B behaved similarly, and the control subject had a greater angle value (17°). However, it is known that the angle during initiation of the stance phase is between 25° and 35° for the hip joint[11, 15, 16)]. The control subject also had a greater extension angle in the stance phase (16.4°), which was close to the toe-off angle, as expected in comparison with other authors[11, 15)]. Patient B showed a rigid flexion and extension mass pattern that can be assumed to be influenced by spasticity. This event happened in both the stance and swing phases with a synergic mass pattern, as described by other authors[15, 17, 18)].

Patient A had their maximum extension angle in the stance phase, which occurred earlier than in the others subjects. As mentioned before, the stance phase of hemiparetic limbs has a shorter duration, and the maximal extension angle may occur earlier in the gait cycle. In the swing phase, patient A showed a greater maximum flexion angle compared with patient B and the control. The excessive hip flexion of patient A during the swing phase could be a form of compensation due to lower knee flexion and was probably also due to the incapacity of the plantar flexor muscles to generate enough power for the initial swing phase. The total sagittal plane excursion was similar in patient A and the control and it could represent a greater limb distance reached during walking. Even though step and stride frequency and distance were not assessed in this study, Mulroy et al.[19)] suggested that larger hip excursion may be due to superior strength in the extensor muscles of the hip.

The behavior of the knee joint angles was similar for patient and the control subject. However, patient B started the stance phase with a higher value at the initial contact angle. Correa et al.[14)] demonstrated that during the stance phase, patients who had had a stroke show co-contractions of agonist and antagonist muscles at the ankle and knee joints and that these adaptations may allow a safer and more stable gait pattern. This may have occurred with patient B, who showed a lower sagittal plane excursion in accordance with the hip angles described before, representing a stiff knee gait pattern.

At the toe-off line, the control participant showed sufficient joint range of motion in comparison with the results of other studies, although the evaluation of joint range of motion also depends on others variables like stride length and velocity. Conversely, patient A showed a compensation adopted in mid stance that reflected the knee position at the end of the stance phase until the maximal flexion angle in the swing phase[11, 16, 20)].

In contrast to the results presented previously, patient A behaved differently compared with the control subject, indicating a higher difference at the initial contact angle. This behavior did not explain any other alterations in the hip and knee curves and seems to be a particular compensatory strategy, which suggests an enhancement in the maximal dorsiflexion angle using kinetic energy, but this was not assessed in this study. On the other hand, the maximal dorsiflexion angle occurred with a satisfactory range of motion for all participants compared with the results reported by other authors but at different times of the gait cycle for the hemiparetic patients in comparison with the control subject[11, 15, 16)]. This is probably happens due to the short stance phase of the hemiplegic limb, as mentioned before.

Kinsela et al.[11)] observed that the sagittal excursion of the hip is facilitated by ankle dorsiflexion in the stance phase and by ankle excursion in the sagittal plane. Both patient A and the control subject may have had greater ankle dorsiflexor and plantar flexor strength than patient B, as described by Mulroy et al[19)]. These authors also proposed that a decrease in ankle dorsiflexion during the stance phase and knee flexion sustained throughout the gait cycle suggest a "stiff knee" pattern; this was seen in the present study in patient B. In the early swing phase, it is possible to deduce that there was decreased activation of the anterior tibialis in both hemiparetic patients who presented inadequate dorsiflexion in mid swing in comparison with the control or a limited range of motion due to muscle shortening or due to spasticity. The patients seem to have an insufficient muscle response to achieve a neutral ankle angle for the next step of the gait (heel strike) compared with the control, and the dorsiflexion improvement in the swing phase may reflect an increased intensity of anterior tibialis muscle[20)].

The use of inertial sensors allowed description of the gait cycle of patients who suffered stroke. The lower limb movement of the two patients with hemiparesis in the sagittal plane could be described and analyzed in comparison with a control subject. Thus, patient A and the control subject had similar gait cycle curves at all joints, while patient B showed a rigid synergic pattern. Limitations of this study were the small number of participants, the interference of magnetometers that needs to be considered during data collection and analysis, and gait assessment only from the perspective of the sagittal plane.

The present study demonstrated differences during walking between patients who had had a stroke in different arteries and a control subject, which may be related to an alternative compensatory strategy. Future studies are needed to explore kinematic data of subjects with different affected brain areas. A key point of the present study was the use of inertial sensors during kinematic gait analysis, which demonstrated a practical and functional approach.

ACKNOWLEDGEMENTS

The authors are thankful for the grants from the Ministry of Science and Technology of Brazil/CNPq (#70/2009) and from the Araucaria Research Foundation/PPSUS (Research Program for the Unified Health System, SESA-PR/MS-Decit/CNPq, #04/2013). The last author wishes to thank CNPq for the Productivity Scholarship (2014-2018). No conflicts of interest were reported for this study.

REFERENCES

1) Woolley SM: Characteristics of gait in hemiplegia. Top Stroke Rehabil, 2001, 7: 1–18. [Medline] [CrossRef]
2) Marigold DS, Patla AE: Adapting locomotion to different surface compliances: neuromuscular responses and changes in movement dynamics. J Neurophysiol, 2005, 94: 1733–1750. [Medline] [CrossRef]
3) Shkuratova N, Morris ME, Huxham F: Effects of age on balance control during walking. Arch Phys Med Rehabil, 2004, 85: 582–588. [Medline] [CrossRef]
4) Garcia RK, Nelson AJ, Ling W, et al.: Comparing stepping-in-place and gait ability in adults with and without hemiplegia. Arch Phys Med Rehabil, 2001, 82: 36–42. [Medline] [CrossRef]
5) Olney SJ, Richards C: Hemiparetic gait following stroke. Part I: Characteristics. Gait Posture, 1996, 4: 136–148. [CrossRef]
6) Levine D, Richards J, Whittle MW: Whittle's Gait Analysis, 5th ed. London: Churchill Livingstone, 2012, p 177.
7) Schepers HM, Roetenberg D, Veltink PH: Ambulatory human motion tracking by fusion of inertial and magnetic sensing with adaptive actuation. Med Biol Eng Comput, 2010, 48: 27–37. [Medline] [CrossRef]
8) Bonato P: Wearable sensors/systems and their impact on biomedical engineering. IEEE Eng Med Biol Mag, 2003, 22: 18–20. [Medline] [CrossRef]
9) Guo Y, Wu D, Liu G, et al.: A low-cost body inertial-sensing network for practical gait discrimination of hemiplegia patients. Telemed J E Health, 2012, 18: 748–754. [Medline] [CrossRef]
10) Carmo AA, Kleiner AF, Costa PH, et al.: Three-dimensional kinematic analysis of upper and lower limb motion during gait of post-stroke patients. Braz J Med Biol Res, 2012, 45: 537–545. [Medline] [CrossRef]
11) Kinsella S, Moran K: Gait pattern categorization of stroke participants with equinus deformity of the foot. Gait Posture, 2008, 27: 144–151. [Medline] [CrossRef]
12) Chaler J, Müller B, Maiques A, et al.: Suspected feigned knee extensor weakness: usefulness of 3D gait analysis. Case report. Gait Posture, 2010, 32: 354–357. [Medline] [CrossRef]
13) Chen CL, Chen HC, Tang SF, et al.: Gait performance with compensatory adaptations in stroke patients with different degrees of motor recovery. Am J Phys Med Rehabil, 2003, 82: 925–935. [Medline] [CrossRef]
14) Corrêa FI, Soares F, Andrade DV, et al.: [Muscle activity during gait following stroke]. Arq Neuropsiquiatr, 2005, 63: 847–851. [Medline]
15) Verma R, Arya KN, Sharma P, et al.: Understanding gait control in post-stroke: implications for management. J Bodyw Mov Ther, 2012, 16: 14–21. [Medline] [CrossRef]
16) Piccinini L, Cimolin V, D'Angelo MG, et al.: 3D gait analysis in patients with hereditary spastic paraparesis and spastic diplegia: a kinematic, kinetic and EMG comparison. Eur J Paediatr Neurol, 2011, 15: 138–145. [Medline] [CrossRef]
17) Neckel ND, Blonien N, Nichols D, et al.: Abnormal joint torque patterns exhibited by chronic stroke subjects while walking with a prescribed physiological gait pattern. J Neuroeng Rehabil, 2008, 5: 19. [Medline] [CrossRef]
18) Titianova EB, Tarkka IM: Asymmetry in walking performance and postural sway in patients with chronic unilateral cerebral infarction. J Rehabil Res Dev, 1995, 32: 236–244. [Medline]
19) Mulroy S, Gronley J, Weiss W, et al.: Use of cluster analysis for gait pattern classification of patients in the early and late recovery phases following stroke. Gait Posture, 2003, 18: 114–125. [Medline] [CrossRef]
20) Arnold AS, Thelen DG, Schwartz MH, et al.: Muscular coordination of knee motion during the terminal-swing phase of normal gait. J Biomech, 2007, 40: 3314–3324. [Medline] [CrossRef]