AQUISITION SYSTEM FOR DATA ANALYSIS OF GAIT WITH LOWER LIMB PROSTHESIS - ANDAR

M. V. Correia[1,2], S. Cunha[2], D. Viegas[1], C. Cunha[2,3], P. Diniz[2,3], E. Mendes[3]
[1]Instituto de Engenharia Biomédica, Laboratório de Sinal e Imagem Biomédica
[2]Universidade do Porto, Faculd. Engenharia, Dept. Eng. Electrotécnica e Computadores
R. Dr. Roberto Frias, 4200-465 Porto, Portugal
[3]Centro de Reabilitação Profissional de Gaia
Avenida João Paulo II, 4410-406 Arcozelo, Portugal
mcorreia@fe.up.pt, emilia.mendes@crpg.pt

Abstract

This paper describes the implementation of a gait analysis device used to measure the use of prosthesis by lower limb amputated subjects. Based on dead-reckoning sensors only (two accelerometers, one gyroscope and pressure sensors, all solid state), it is used to compute the travelled distance, number of steps, motion speed, leg aperture and related parameters from which assessment of comfort and patient adaptation to the prosthesis can be derived. The core of the real-time data processing set of algorithms is a Kalman filter, as here described, together with some results.

Keywords
Inertial sensors, Kalman filter, gait analysis.

1. INTRODUCTION

In current day rehabilitation practice, prosthesis of lower limbs are custom made, adjusted and analysed within laboratory facilities, where its structural and functional assessment is performed. Yet, from the moment the patient leaves the rehabilitation centre's lab, no record is made of the use or utility of the prosthesis or of its performance. The follow up of the patient's evolution is done with periodic visits to the rehabilitation centre, but without any objective information regarding the effective use of the prosthesis, such as the amount of time it was used, in what circumstances, what was the distance travelled, or how much effort is required from the patient on an everyday basis.

There is a clear need in prosthetics and orthotics practice to perform objective measurements of patient activity over a continuous time span, with minimal disturbance of daily routine [1]. Several systems exist for gait analysis, ranging from low cost ultrasonic apparatus [2] to very high cost setups with several cameras [3]. A common characteristic of those systems, however, is that they are adequate only for operation in the laboratory and for short periods of time.

This paper describes the work done in a research project with the aim to study and develop methods and means to collect and analyse gait data, which will allow the assessment of effective use and utility of lower limb prosthesis on the everyday life of an amputated person, at its domestic or working environments. It will provide data to conclude how well adapted the prosthesis is and what advantage such investment provides to patients at different ages.

An electronic device was developed so that it can be attached to a prosthesis and operate from a battery power supply for a period of at least seven days. The device includes a set of inertial sensors and electronic components to perform data recording during normal day-by-day activity of the patient, to be later analysed at the rehabilitation centre. A similar device has been developed by Benbasat *et al.*[4] that includes a complete inertial unit with six degrees of freedom and several pressure sensors but with a different purpose and much higher cost.

The paper is organized as follows. Section 2 briefly describes the human gait cycle. The signal acquisition and processing are described in sections 3 and 4, respectively. Finally, results and conclusions are presented in sections 5 and 6.

2. GAIT CYCLE

The human gait is a natural form of human locomotion that can be described by several kinetic or kinematic characteristics. It is often modelled as a harmonic motion similar to a pendulum. This type of biological motion contains plenty of information, making it a very complex and unique activity. Despite this complexity, there are common patterns that can be identified in human gait. The normal gait cycle is divided into two phases, namely, stance and swing. The objective of stance is to provide support, stability and propulsion and the swing phase provides ground clearance and limb advancement. Knee flexion is essential during swing to lift the foot off the ground for limb advancement and hip translation. Human gait is illustrated in Figure 1. Gait analysis was essential for choosing the adequate sensors for our purposes.

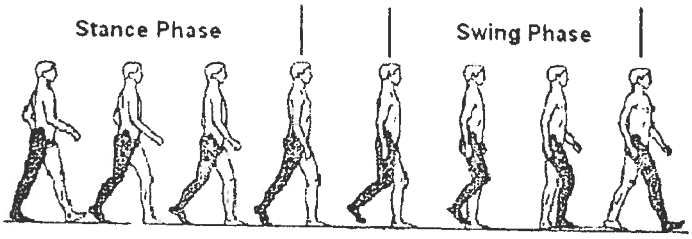


Figure 1 – The gait cycle

3. SENSORS AND SIGNAL ACQUISITION

The device developed within this project employs micro-electromechanical inertial sensors (MEMS), i.e., miniature gyroscopes and accelerometers and piezoresistive sensors placed insole. The measurements include the number of steps, pressures on the foot during the gait cycle as well as linear accelerations and angular speed of the foot, which upon integration, estimates walked distance. Figure 1 shows the prototype developed for test purposes.

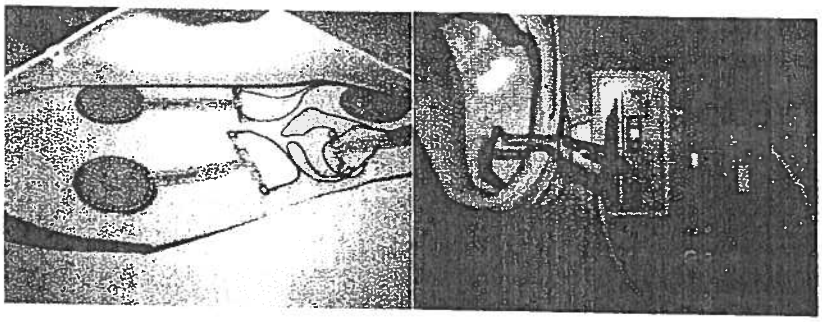


Figure 1 – The device prototype with pressure sensing insole, inertial sensors board and processing board.

The sensors used were force sensing resistors (FSR) and low cost, solid state inertial sensors, such as accelerometers and gyroscope. Accelerometers axes are placed parallel to the sagital plane of the body, giving the accelerations on two orthogonal axes on this plane and the angular speed is given by gyroscope perpendicularly to it.

4. SIGNAL PROCESSING AND DATA ANALYSIS

The acquired signals were computed with an Extended Kalman filter structure [5], which is an optimal recursive data processing algorithm, especially adapted and implemented in software for this purpose. This filter, using a discrete time model and discrete observations, was designed to run in small microcontrollers and perform in the order of 100 cycles per second in real-time. This required optimization of the chosen state and observations vectors. The direct inputs for the filter are accelerations and angular speed measurement. The observations derive from the properties that the foot is not moving during the stance phase.

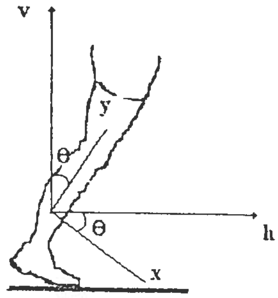


Figure 2 – Referentials in the physical system model

The signals are acquired with respect to the referential in the sagital plane of the moving body (Figure 2). So, the accelerations in the inertial referential (horizontal and vertical axis with respect to local Earth) can be written as:

$$a_h = a_x \cos\theta + a_y \sin\theta \tag{1}$$

$$a_v = -a_x \sin\theta + a_y \cos\theta \tag{2}$$

where a_x and a_y are the accelerations measured by the accelerometers and θ is the integration of the angle rate measured by the gyro.

Speed is obtained by integration of the accelerations in the inertial referential:

$$v_h = \int \left(a_x \cos\theta + a_y \sin\theta\right) dt \tag{3}$$

$$v_v = \int \left(-a_x \sin\theta + a_y \cos\theta\right) dt \tag{4}$$

where v_h is the horizontal speed, and v_v the vertical speed.

The above integration will drift due to accelerometer and gyro measurement bias and noise. Therefore, vertical accelerometer bias (O_{av}) and the integrated angle error ($\Delta\theta$) were added to the state vector, which is:

$$x_{(k)} = \left[\Delta\theta_{(k)} \; O_{av(k)} \; v_{h(k)} \; v_{v(k)}\right] \tag{5}$$

The angle error absorbs the effect of the gyro drift, errors in the integration due to same motion outside the sagital plane, and the horizontal effect of the accelerometer's offset. Since $\Delta\theta$ includes such effect, it would be redundant include another state specifically for this offset. Furthermore, it would lead to observability problems with such low cost devices. The vertical and horizontal speeds were included in the state vector because they were the observed parameters, during the stance phase of the gait cycle. This phase was detected using data from the FSR sensors.

The dynamics of the system can be described as follows:

$$\begin{cases} \Delta\theta_{(k+1)} = \Delta\theta_{(k)} \\ O_{av(k+1)} = O_{av(k)} \\ v_{h(k+1)} = v_{h(k)} + \left[a_{x(k)} \cos\left(\theta + \Delta\theta_{(k)}\right) + \left(a_{y(k)} + O_{av(k)}\right)\sin\left(\theta + \Delta\theta_{(k)}\right)\right] \cdot dt \\ v_{v(k+1)} = v_{v(k)} - \left[a_{x(k)} \sin\left(\theta + \Delta\theta_{(k)}\right) - \left(a_{y(k)} + O_{av(k)}\right)\cos\left(\theta + \Delta\theta_{(k)}\right)\right] \cdot dt \end{cases} \tag{6}$$

The horizontal and vertical positions result from the integration of the speed in each direction.

In this study, the aim was to detect the behaviour of the patient with lower limb prosthesis on everyday life, so a direct outcome of the filter is the accurate measurement of travelled distance, as an indicator of the patient activity.

5. RESULTS

Figure 3 shows the different signals acquired from the inertial sensors and the data processing by the filter structure. The angular speed and accelerations are sensed directly. The others variable are computed by the filter. The individual cycles can be clearly identified in the picture.

Figure 4 describes the gait from analysing the angle, the position and the pressure data from the sensors. The stance phase corresponds to the period where there is high foot pressure and angle increase, while the swing phase is indicated by the angle decrease and no pressure on the foot. Also, the hip and, consequently body, translate during the swing phase with an increase in position.

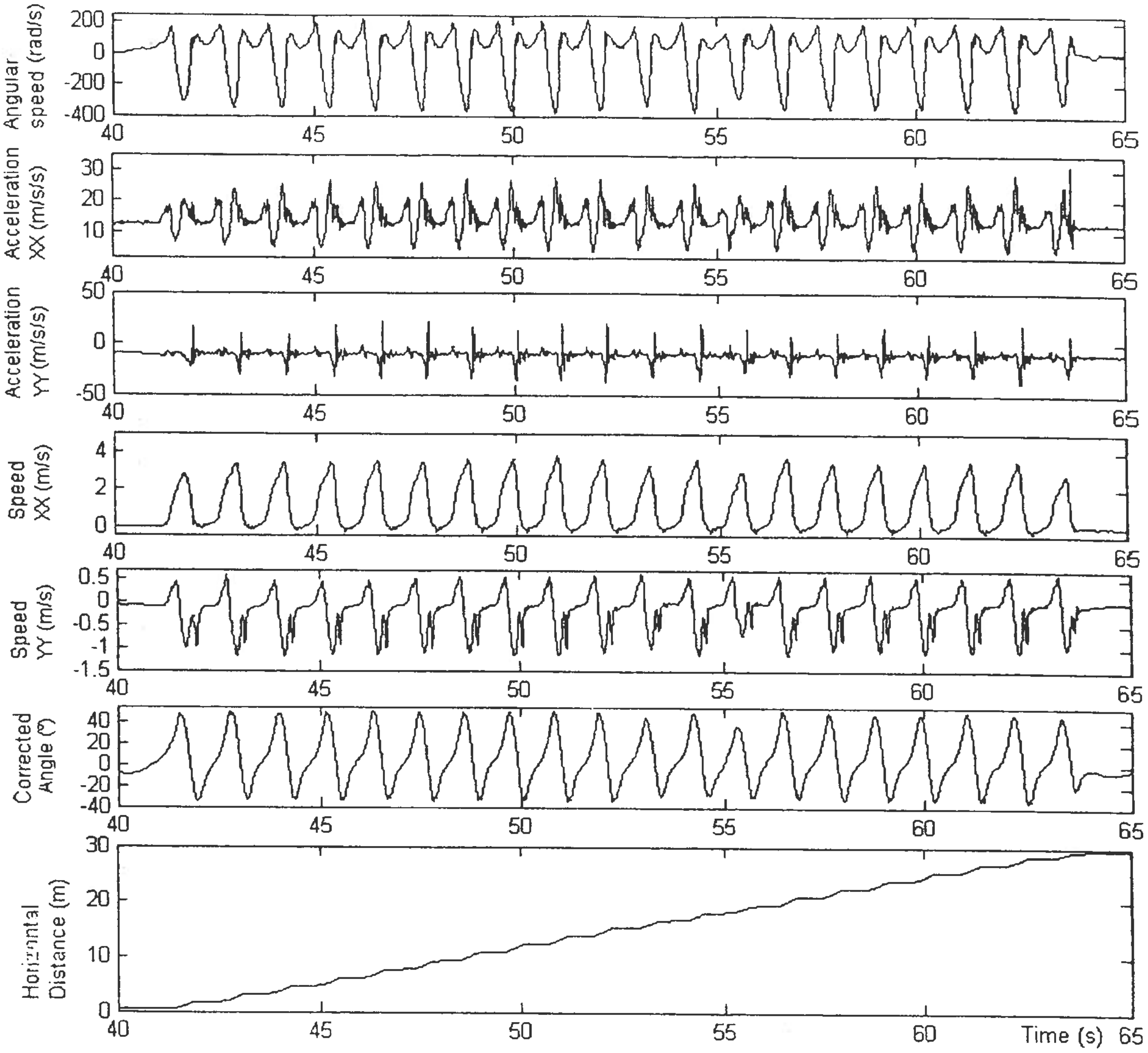


Figure 3 – Kinematic signals acquired by the sensors and computed by the Kalman filter.

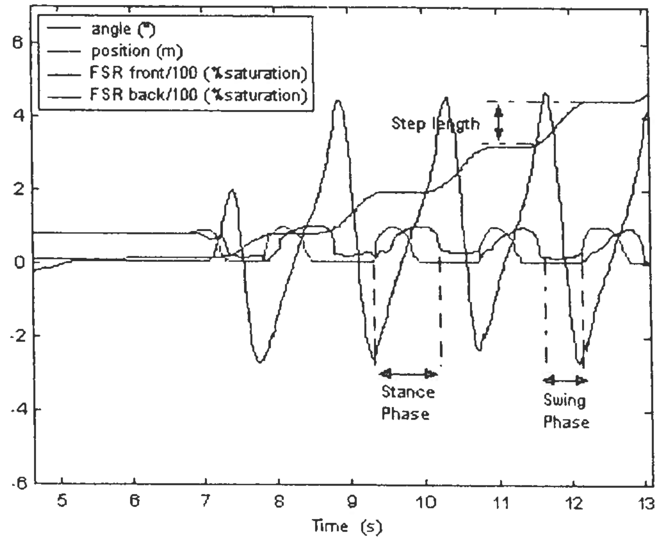


Figure 4 – Phases of the gait cycle as detected by the sensors

Table 1 shows the ground-truth values and the results obtained with the prototype of locomotion in a series of segments with a person 1.83m high and a weight of 76Kg.

Table 1 – Values of locomotion

Steps	Ground-truth distance (m)	Distance (m)	Maximum speed (m/s)	Medium step length
21	29	30,0	3,956	1,399
20	29	29,9	3,938	1,480
20	29	28,6	3,848	1,373
19	29	28,8	4,202	1,549
20	29	27,9	3,979	1,474
21	29	29,2	3,852	1,342
16	29	29,7	4,264	1,711
20	29	29,8	3,985	1,565
21	29	25,4	3,342	1,161
19	29	28,9	3,782	1,479
198	290	288,2		

As we can see, results are very accurate. Additionally, there are other measurements, such as the step length and travel speed, that can be useful for gait analysis. After many experiments under different conditions, the obtained results present an error consistently below 5%, as those shown in Table 1.

6. CONCLUSIONS

In spite of the simplicity of the employed hardware and its low cost, the obtained results are very satisfactory, providing much richer information than any other of the few devices available for the same purpose in the market. No other low cost device provided accurate travelled distance estimates. Furthermore, the exclusive use of dead-reckoning devices that do not require patient specific calibration or any interaction with the surrounding environment proved to be appropriate when minimal interaction with the patient day-to-day life is of key importance. The developed prototype is now being further miniaturized and prepared for construction in quantities to be used in an extensive test conducted by the rehabilitation centre.

Acknowledgments

The authors would like to acknowledge the financial support of Fundação Ilídio Pinho and Universidade do Porto given by their programme of Scientific Research in the Undergraduate Level.

References

[1] Sykes, L., Ross, E.R.S, Powell, E.S., and Edwards, J., "Objective measurement of use of the reciprocating gait orthosis (RGO) and the electrically augmented RGO in adult patients with spinal cord lesions", *Prosthetics and Orthotics International,* vol. 20, 1996, pp. 182-190.

[2] Huitema, R.B., Hof, A. L., Postema, K., "Ultrasonic motion analysis system–measurement of temporal and spatial gait parameters", *Journal of Biomechanics,* Elsevier, vol.35, 2002, pp. 837-842.

[3] URL: *http://www.simi.de.*

[4] Benbasat, A.Y., Morris, S.J., Paradiso, J.A., "A wireless modular sensor architecture and its application in on-shoe gait analysis", In proceedings of *IEEE Sensors Conference,* Toronto, Ontario, Canada, October 2003.

[5] Gelb, A., Applied Optimal Estimation, MIT Press, Cambridge, Massachussets, USA, 1974.

AQUISITION SYSTEM FOR DATA ANALYSIS OF GAIT WITH LOWER LIMB PROSTHESIS - ANDAR

M. V. Correia[1,2], S. Cunha[2], D. Viegas[1], C. Cunha[2,3], P. Diniz[2,3], E. Mendes[3]
[1]Instituto de Engenharia Biomédica, Laboratório de Sinal e Imagem Biomédica
[2]Universidade do Porto, Faculd. Engenharia, Dept. Eng. Electrotécnica e Computadores
R. Dr. Roberto Frias, 4200-465 Porto, Portugal
[3]Centro de Reabilitação Profissional de Gaia
Avenida João Paulo II, 4410-406 Arcozelo, Portugal
mcorreia@fe.up.pt, emilia.mendes@crpg.pt

Abstract

This paper describes the implementation of a gait analysis device used to measure the use of prosthesis by lower limb amputated subjects. Based on dead-reckoning sensors only (two accelerometers, one gyroscope and pressure sensors, all solid state), it is used to compute the travelled distance, number of steps, motion speed, leg aperture and related parameters from which assessment of comfort and patient adaptation to the prosthesis can be derived. The core of the real-time data processing set of algorithms is a Kalman filter, as here described, together with some results.

Keywords
Inertial sensors, Kalman filter, gait analysis.

1. INTRODUCTION

In current day rehabilitation practice, prosthesis of lower limbs are custom made, adjusted and analysed within laboratory facilities, where its structural and functional assessment is performed. Yet, from the moment the patient leaves the rehabilitation centre's lab, no record is made of the use or utility of the prosthesis or of its performance. The follow up of the patient's evolution is done with periodic visits to the rehabilitation centre, but without any objective information regarding the effective use of the prosthesis, such as the amount of time it was used, in what circumstances, what was the distance travelled, or how much effort is required from the patient on an everyday basis.

There is a clear need in prosthetics and orthotics practice to perform objective measurements of patient activity over a continuous time span, with minimal disturbance of daily routine [1]. Several systems exist for gait analysis, ranging from low cost ultrasonic apparatus [2] to very high cost setups with several cameras [3]. A common characteristic of those systems, however, is that they are adequate only for operation in the laboratory and for short periods of time.

This paper describes the work done in a research project with the aim to study and develop methods and means to collect and analyse gait data, which will allow the assessment of effective use and utility of lower limb prosthesis on the everyday life of an amputated person, at its domestic or working environments. It will provide data to conclude how well adapted the prosthesis is and what advantage such investment provides to patients at different ages.

An electronic device was developed so that it can be attached to a prosthesis and operate from a battery power supply for a period of at least seven days. The device includes a set of inertial sensors and electronic components to perform data recording during normal day-by-day activity of the patient, to be later analysed at the rehabilitation centre. A similar device has been developed by Benbasat *et al.*[4] that includes a complete inertial unit with six degrees of freedom and several pressure sensors but with a different purpose and much higher cost.

The paper is organized as follows. Section 2 briefly describes the human gait cycle. The signal acquisition and processing are described in sections 3 and 4, respectively. Finally, results and conclusions are presented in sections 5 and 6.

2. GAIT CYCLE

The human gait is a natural form of human locomotion that can be described by several kinetic or kinematic characteristics. It is often modelled as a harmonic motion similar to a pendulum. This type of biological motion contains plenty of information, making it a very complex and unique activity. Despite this complexity, there are common patterns that can be identified in human gait. The normal gait cycle is divided into two phases, namely, stance and swing. The objective of stance is to provide support, stability and propulsion and the swing phase provides ground clearance and limb advancement. Knee flexion is essential during swing to lift the foot off the ground for limb advancement and hip translation. Human gait is illustrated in Figure 1. Gait analysis was essential for choosing the adequate sensors for our purposes.

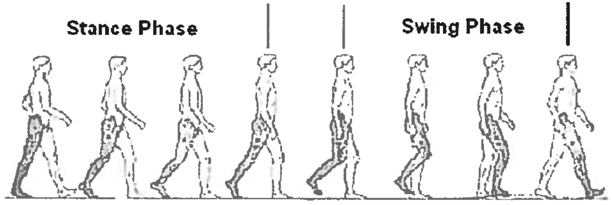


Figure 1 – The gait cycle

3. SENSORS AND SIGNAL ACQUISITION

The device developed within this project employs micro-electromechanical inertial sensors (MEMS), i.e., miniature gyroscopes and accelerometers and piezoresistive sensors placed insole. The measurements include the number of steps, pressures on the foot during the gait cycle as well as linear accelerations and angular speed of the foot, which upon integration, estimates walked distance. Figure 2 shows the prototype developed for test purposes.

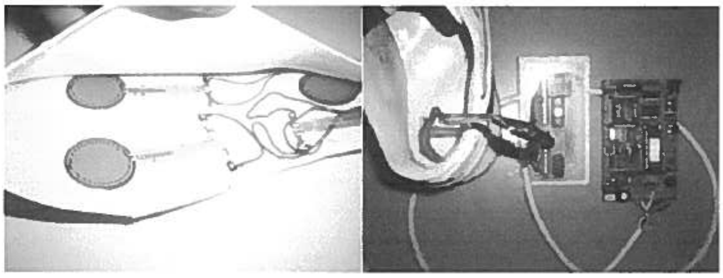


Figure 2 – The device prototype with pressure sensing insole, inertial sensors board and processing board.

The sensors used were force sensing resistors (FSR) and low cost, solid state inertial sensors, such as accelerometers and gyroscope. Accelerometers axes are placed parallel to the sagital plane of the body, giving the accelerations on two orthogonal axes on this plane and the angular speed is given by gyroscope perpendicularly to it.

4. SIGNAL PROCESSING AND DATA ANALYSIS

The acquired signals were computed with an Extended Kalman filter structure [5], which is an optimal recursive data processing algorithm, especially adapted and implemented in software for this purpose. This filter, using a discrete time model and discrete observations, was designed to run in small microcontrollers and perform in the order of 100 cycles per second in real-time. This required optimization of the chosen state and observations vectors. The direct inputs for the filter are accelerations and angular speed measurement. The observations derive from the properties that the foot is not moving during the stance phase.

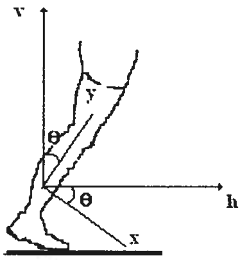


Figure 3 – Referentials in the physical system model

The signals are acquired with respect to the referential in the sagital plane of the moving body (Figure 3). So, the accelerations in the inertial referential (horizontal and vertical axis with respect to local Earth) can be written as:

$$a_h = a_x \cos\theta + a_y \sin\theta \tag{1}$$

$$a_v = -a_x \sin\theta + a_y \cos\theta \tag{2}$$

where a_x and a_y are the accelerations measured by the accelerometers and θ is the integration of the angle rate measured by the gyro.

Speed is obtained by integration of the accelerations in the inertial referential:

$$v_h = \int (a_x \cos\theta + a_y \sin\theta) dt \tag{3}$$

$$v_v = \int (-a_x \sin\theta + a_y \cos\theta) dt \tag{4}$$

where v_h is the horizontal speed, and v_v the vertical speed.

The above integration will drift due to accelerometer and gyro measurement bias and noise. Therefore, vertical accelerometer bias (O_{av}) and the integrated angle error ($\Delta\theta$) were added to the state vector, which is:

$$x_{(k)} = \left[\Delta\theta_{(k)} \; O_{av(k)} \; v_{h(k)} \; v_{v(k)}\right] \tag{5}$$

The angle error absorbs the effect of the gyro drift, errors in the integration due to same motion outside the sagital plane, and the horizontal effect of the accelerometer's offset. Since $\Delta\theta$ includes such effect, it would be redundant include another state specifically for this offset. Furthermore, it would lead to observability problems with such low cost devices. The vertical and horizontal speeds were included in the state vector because they were the observed parameters, during the stance phase of the gait cycle. This phase was detected using data from the FSR sensors.

The dynamics of the system can be described as follows:

$$\begin{cases} \Delta\theta_{(k+1)} = \Delta\theta_{(k)} \\ O_{av(k+1)} = O_{av(k)} \\ v_{h(k+1)} = v_{h(k)} + \left[a_{x(k)} \cos\left(\theta + \Delta\theta_{(k)}\right) + \left(a_{y(k)} + O_{av(k)}\right)\sin\left(\theta + \Delta\theta_{(k)}\right)\right] \cdot dt \\ v_{v(k+1)} = v_{v(k)} - \left[a_{x(k)} \sin\left(\theta + \Delta\theta_{(k)}\right) - \left(a_{y(k)} + O_{av(k)}\right)\cos\left(\theta + \Delta\theta_{(k)}\right)\right] \cdot dt \end{cases} \tag{6}$$

The horizontal and vertical positions result from the integration of the speed in each direction.

In this study, the aim was to detect the behaviour of the patient with lower limb prosthesis on everyday life, so a direct outcome of the filter is the accurate measurement of travelled distance, as an indicator of the patient activity.

5. RESULTS

Figure 4 shows the different signals acquired from the inertial sensors and the data processing by the filter structure. The angular speed and accelerations are sensed directly. The others variable are computed by the filter. The individual cycles can be clearly identified in the picture.

Figure 5 describes the gait from analysing the angle, the position and the pressure data from the sensors. The stance phase corresponds to the period where there is high foot pressure and angle increase, while the swing phase is indicated by the angle decrease and no pressure on the foot. Also, the hip and, consequently body, translate during the swing phase with an increase in position.

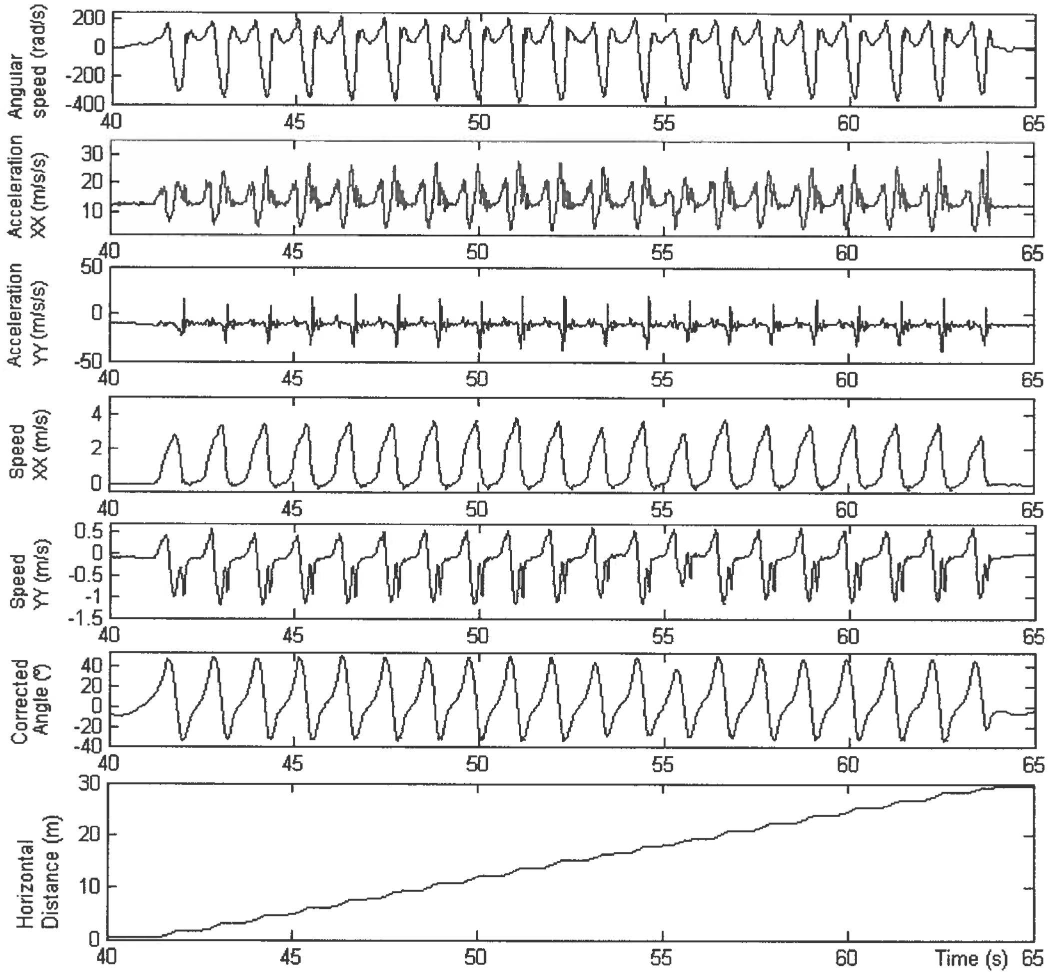


Figure 4 – Kinematic signals acquired by the sensors and computed by the Kalman filter.

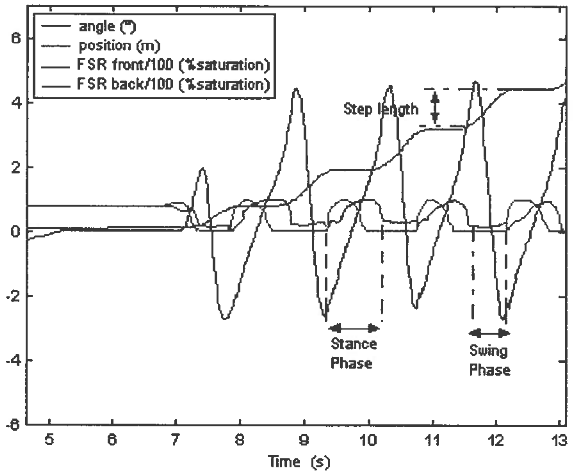


Figure 5 – Phases of the gait cycle as detected by the sensors

Table 1 shows the ground-truth values and the results obtained with the prototype of locomotion in a series of segments with a person 1.83m high and a weight of 76Kg.

Table 1 – Values of locomotion

Steps	Ground-truth distance (m)	Distance (m)	Maximum speed (m/s)	Medium step length
21	29	30,0	3,956	1,399
20	29	29,9	3,938	1,480
20	29	28,6	3,848	1,373
19	29	28,8	4,202	1,549
20	29	27,9	3,979	1,474
21	29	29,2	3,852	1,342
16	29	29,7	4,264	1,711
20	29	29,8	3,985	1,565
21	29	25,4	3,342	1,161
19	29	28,9	3,782	1,479
198	290	288,2		

As we can see, results are very accurate. Additionally, there are other measurements, such as the step length and travel speed, that can be useful for gait analysis. After many experiments under different conditions, the obtained results present an error consistently below 5%, as those shown in Table 1.

6. CONCLUSIONS

In spite of the simplicity of the employed hardware and its low cost, the obtained results are very satisfactory, providing much richer information than any other of the few devices available for the same purpose in the market. No other low cost device provided accurate travelled distance estimates. Furthermore, the exclusive use of dead-reckoning devices that do not require patient specific calibration or any interaction with the surrounding environment proved to be appropriate when minimal interaction with the patient day-to-day life is of key importance. The developed prototype is now being further miniaturized and prepared for construction in quantities to be used in an extensive test conducted by the rehabilitation centre.

Acknowledgments

The authors would like to acknowledge the financial support of Fundação Ilídio Pinho and Universidade do Porto given by their programme of Scientific Research in the Undergraduate Level.

References

[1] Sykes, L., Ross, E.R.S, Powell, E.S., and Edwards, J., "Objective measurement of use of the reciprocating gait orthosis (RGO) and the electrically augmented RGO in adult patients with spinal cord lesions", *Prosthetics and Orthotics International,* vol. 20, 1996, pp. 182-190.

[2] Huitema, R.B., Hof, A. L., Postema, K., "Ultrasonic motion analysis system–measurement of temporal and spatial gait parameters", *Journal of Biomechanics,* Elsevier, vol.35, 2002, pp. 837-842.

[3] URL: *http://www.simi.de.*

[4] Benbasat, A.Y., Morris, S.J., Paradiso, J.A., "A wireless modular sensor architecture and its application in on-shoe gait analysis", In proceedings of *IEEE Sensors Conference,* Toronto, Ontario, Canada, October 2003.

[5] Gelb, A., Applied Optimal Estimation, MIT Press, Cambridge, Massachussets, USA, 1974.

AQUISITION SYSTEM FOR DATA ANALYSIS OF GAIT WITH LOWER LIMB PROSTHESIS - ANDAR

M. V. Correia[1,2], S. Cunha[2], D. Viegas[1], C. Cunha[2,3], P. Diniz[2,3], E. Mendes[3]
[1]Instituto de Engenharia Biomédica, Laboratório de Sinal e Imagem Biomédica
[2]Universidade do Porto, Faculd. Engenharia, Dept. Eng. Electrotécnica e Computadores
R. Dr. Roberto Frias, 4200-465 Porto, Portugal
[3]Centro de Reabilitação Profissional de Gaia
Avenida João Paulo II, 4410-406 Arcozelo, Portugal
mcorreia@fe.up.pt, emilia.mendes@crpg.pt

Abstract

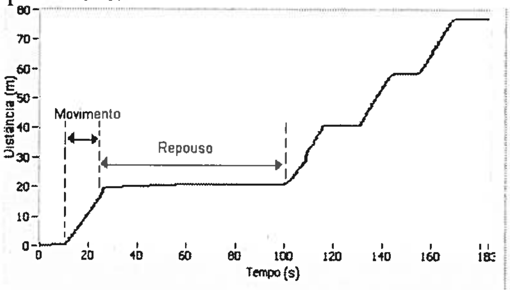
In current day rehabilitation practice, prosthesis of lower limbs are custom made, adjusted and analysed within laboratory facilities, where its structural and functional assessment is performed. Yet, from the moment the patient leaves the rehabilitation centre's lab, no record is made of the use or utility of the prosthesis or of its performance. The follow up of the patient's evolution is done with periodic visits to the rehabilitation centre, but without any objective information regarding the effective use of the prosthesis, such as the amount of time it was used, in what circumstances, what was the distance travelled, or how much effort is required from the patient on an everyday basis.

This paper describes the work done in a research project with the aim to study and develop methods and means to collect and analyse gait data, which will allow the assessment of effective use and utility of lower limb prosthesis on the everyday life of an amputated person, at its domestic or working environments. It will provide data to conclude how well adapted the prosthesis is and what advantage such investment provides to patients at different ages.

An electronic device was developed so that it can be attached to a prosthesis and operate from a battery power supply for a period of at least seven days. The device includes a set of sensors and electronic components to perform data recording during normal day-by-day activity of the patient, to be later analysed at the rehabilitation centre.

The device employs micro-electromechanical inertial sensors (MEMS), i.e., miniature gyroscopes and accelerometers and force sensing resistors placed insole. The measurements include the number of steps, pressures on the foot during the gait cycle as well as linear accelerations and angular speed of the foot. The distance travelled is computed with a Kalman filter structure, specially adapted and implemented in software for this purpose. It takes advantage of the cyclic movement of human gait in order to obtain the best estimate of the velocity and orientation of the foot in the sagital plane, successfully correcting drifts and offsets of the inertial sensors, with very significant gains in measurement accuracy.

In figure 1 we show the increase of the computed travelled distance for a period of 3 minutes, up to a total of approximately 75 meters. The path was covered in four segments of 20 meters each, as can be seen by the moments of walking and rest in the graph. After several trials, the distance results show errors below 5%, which exceeded initial expectations.



[Computed travelled distance for a path of 80 meters.]

An initial prototype is already developed and functional has a proof-of-concept. Additional work is being carried out in order to make it smaller, more robust and reliable for everyday use, and to perform its validation through field tests. Financial support was given by Fundação Ilídio Pinho and University of Porto by a grant from a programme to foster scientific research at the undergraduate level.

PROCEEDINGS IADAT-aci2005

IADAT

IADAT International Conference on

Automation, Control and Instrumentation

Focused on

Innovation, Technology and Research

2-4 February 2005
Bilbao, SPAIN

Edited by
IADAT