Detection of Gait Events and Assessment of Fall Risk Using Accelerometers in Assisted Gait

A. Tereso[1], M. Martins[1], C. P. Santos[1], M. Vieira da Silva [2], L. Gonçalves[1] and L. Rocha[1]
[1]*Industrial Electronics Dep., Minho University, Guimarães, Portugal*
[2]*Braga Hospital, Braga, Portugal*
anamftereso@gmail.com, {mariam, cristina, luis.goncalves, lrocha}@dei.uminho.pt,
vieiradasilva.hospitaldebraga@gmail.com

Keywords: Walker, Accelerometer, Assisted-Gait, Stability, Fall Risk, KOA (Knee Osteoarthritis), TKA (Total Knee Arthroplasty).

Abstract: The use of the walker in rehabilitation has increased in the past few years. Therapists evaluate patient's rehabilitation by observation and subjective tests. Thus, it is necessary the use of an assistive tool which can measure and quantify the patient's walker-assisted movement and stability, providing an objective clinical assessment. The aim of this study is to detect differences in assisted gait when using the assistive devices (ADs) – crutches, standard walker and rollator (4-wheeled walker) with forearm supports (RFS) - in patients with knee osteoarthritis (KOA) that suffered the surgery - Total Knee Arthroplasty (TKA). Additionally, it is to verify the link between gait parameters and acceleration signals.

The evaluation is reached by the use of two 3 axis-accelerometers. The signals extracted from the sensors, at the ankle and trunk, are related to gait events and evaluation of fall risk, respectively. Results show that despite the differences between the signals obtained with the three ADs and with the subjects in this study, it is possible to identify effectively the gait parameters and prove the stability that the RFS provides.

1 INTRODUCTION

Pain relief and the improvement of knee function are the two main reasons for total knee arthroplasty (TKA) in cases of Knee Osteoarthritis (KOA). KOA patients suffer by pain, stiffness and decreased range of motion of the knee, which provokes the reduction of their mobility (Kaufman et al., 2001).

Precise motor function evaluation in rehabilitation programs is a major challenge in clinical practice and has gained widespread interest with recent technologies. Nowadays, in assistive device rehabilitation, therapists evaluate patient's rehabilitation by observation and subjective tests. Such information is qualitative and final clinical decisions are strongly empirical and subjective. This evaluation can be more objective and quantitative, if it applies gait techniques that allow a systematic study and characterization of the human locomotion like accelerometers - low cost wearable sensor systems. These devices are easy to use, can be positioned closed to the places that are supposed to be, are portable and have several biomedical applications (Watanabe et al., 2011).

In this study, it is proposed to assess gait parameters (stance, swing, stride time, etc.) and their variability in assisted gait with three different assistive devices (ADs) (crutches, standard walker and rollator with forearm supports (RFS)) with KOA patients that suffered the surgery TKA. This evaluation was reached with two accelerometers placed at the ankle to detect gait events (toe-off and heel strike) and at the trunk to assess the centre of mass (COM) displacement of the subject. These will provide information about the stability provided by the ADs, as well as estimation of fall risk. The choice of the spatiotemporal (stride, swing and stance time, velocity, cadence and step length) over the kinematics parameters, it was because these parameters provide an objective measurement tool and can help in evaluating KOA severity, effectiveness of treatment and might help in disease management (Debi et al. 2011). It was not possible to compare in this study the assisted gait with unassisted gait, because at the moment of this evaluation, the patients were in recovery from the surgery (between 3rd and 5th day after surgery), so they only could walk with the help of ADs. The

KOA patients are characterized by slow speed, shorter step length and shorter single limb support (Debi et al. 2011). Nowadays, the recovery of KOA patients is made with the help of crutches. However, this type of AD provides an unnatural gait performance and the patients cannot alleviate their pain while walking. Thus, it is intended on this study to find a better solution for the recovery of KOA patients, providing a better gait performance in terms of cadence, speed, comfort and safety. The authors hypothesized that the RFS is a better solution.

The localization of the accelerometer depends on the purpose of the study in terms of gait assessment. In (Sabatini et al. 2005; Doheny et al. 2012), they only used 1 accelerometer positioned at the centre of the foot of the subject to identify gait parameters, but the signal had too much noise and was very irregular. Also, the sensor can be located at the trunk and at the ankle (Lee et al. 2010) for gait evaluation. However, to detect gait events (toe-off and heel strike) it is preferable to place the sensor at the ankle, over the trunk, since it is more sensitive to changes on the lower limbs, providing more information about gait events (Lee et al. 2010). In order to detect automatically such events, the selected method of this study will be based on (Lee et al. 2010). Thus, assisted-gait evaluation with an accelerometer still remains to be validated and in this study it will be performed with the sensor at the ankle. In this case, since the subjects have the knee injured, it would make sense the placement of the sensor at the knee, to better assessment. However, in this study we intended to evaluate the gait and detect gait events, so, since the magnitude of acceleration increases from the head to the ankle (Mathie et al. 2004), the signal in the ankle will be more precise and reliable.

To assess the risk of fall, a sensor should be located near the COM since it is the best place to evaluate with accuracy (Vaughan et al. 1999). Thus, in this work, an accelerometer is placed at the level of the trunk (sacrum), closed to the COM to evaluate its displacement. The assessed COM displacement parameters are based in (Doheny et al. 2012). However, the evaluation performed in (Doheny et al. 2012) was done for the standing position and not during walk. Thus, this study aims to verify the potential of using an accelerometer placed on the trunk to assess fall risk in assisted gait.

Overall, the goal of this study is to detect differences between three ADs in assisted gait by analysing which parameters are most affected by the use of the crutches, standard walker and RFS in patients with KOA, considering gait events and the trunk parameters. As far as the authors know, there are no references on the use of accelerometers in assisted gait with any ADs, only in non-assisted gait. Furthermore, it is intended to verify and validate if the data extracted from the accelerometers is able to detect gait events and changes on the variability of the parameters in assisted gait. The authors expect that the RFS produce the most stable and less variable gait, because of the support provided by the forearms, relatively to the others ADs.

The article is organized as follows. In section 2 it is presented the algorithm, the processing implemented and the parameters analysed. Section 3 briefly reveals the results that were obtained. Sections 4 and 5 are referenced to the discussion and conclusions of these results, respectively.

2 METHODS

2.1 Subjects

A group (N=7) of subjects (3 men and 4 women) aged 67.3 ± 5.06 years that were diagnosed with KOA and suffered the surgery TKA were selected to the trials. The study was conducted at Hospital of Braga, approved by the Ethical Committee, and all the patients signed the informed consent. All trials were filmed with a video camera.

2.2 Test Procedure

In order to assess the effect of the ADs on gait, tests are conducted using crutches, standard walker and RFS (ASBGO walker developed by the authors' team). All the ADs are shown in Figure 1. In these tests, subjects had to walk approximately 10m with the ADs, along a corridor. 3 walking trials for each subject and AD are realized. Then, the mean and standard deviation are estimated for each gait parameter. For each patient the height of the Ads is adjusted. To measure the accelerations of the lower limb and the trunk, two inertial sensors are used. These sensors (SMI, MP6000 of *InvenSense,* which include an accelerometer and a gyroscope, both of them are 3-axial) need a computer and a base station (CC2530 of *Texas Instrument*).

In this study only the accelerometer is used. Two sensors are used, for simplicity, attached to the ankle of the leg with the injured knee and at the sacrum (trunk). The used system configuration and the coordinates of reference for the ankle and trunk are shown in Figure 2. The x-axis, y-axis and z-axis correspond to the medio-lateral (ML), vertical (V) and anterior-posterior (AP) accelerations, respectively.

Figure 1: ADs used in this study. Left image: Crutches; centre image: standard walker; and right image: RFS ASBGo walker).

Figure 2: Description of the axis of the accelerometer at the right ankle (left image) and at the trunk (right image).

2.3 Data Acquisition and Processing

2.3.1 Detection of Gait Parameters

The algorithm implemented in this study for the detection of gait events (heel strike and toe-off) is based on (Lee et al. 2010). These two events are essential for the calculation of gait parameters like stance and swing phase.

The implementation consists on the detection of the time peak of Heel Strike (HS) and Toe-Off (TO) events. First, at each instant of time, the data of each axis is summed and transformed to produce the 'Signal Vector Machine', represented by s:

$$s = \sqrt{a_x^2 + a_y^2 + a_z^2} \quad (1)$$

Where a_x, a_y and a_z are the ML, V and AP accelerations, respectively.

This step is applied since acceleration is highly influenced by the position of the sensor and the 3 axis have significant information. Second, *s* is filtered by a low pass filter (f_{pass}=6Hz, f_{stop}=10Hz) to extract features related to the gait cycle,

$$y[n] = \sum_{i=0}^{10} b_i s[n], \quad (2)$$

where b_i corresponds to the coefficients of the filter. These coefficients are obtained by running the *fdatools* interface in MATLAB. Third, a least-square polynomial derivative approximation filter eliminates noise (points that could be considered wrongly as peaks),

$$z[n] = \frac{1}{10}(1y[n] + y[n-1] - y[n-3] - 2y[n-4]) \quad (3)$$

After this processing, the final step consists on the peak detection. For each gait cycle there are two peaks, each of them corresponding to a gait event (HS and TO). Before this last step, it was necessary to remove some sample points from the start and end of each test, which correspond to the period of acceleration and deceleration in gait, respectively. the duration of these periods is irregular. To validate the detection of such events, it was used one FSR (Force Sensitive Resistor) under the right heel, attached to the shoe, to measure these events.

After detecting both events, some gait parameters can be calculated. The gait cycle is divided in two phases – stance and swing. The majority of the gait cycle is spent in stance phase (60%) and the rest in swing phase (40%). The stance phase corresponds to the moment that the foot is in contact with the ground. The swing phase is the period during which the leg is out of the ground, moving to the next strike. Thus, stance phase begins with HS event and finishes with TO event. Swing phase begins with TO and finishes with HS. Once stance and swing phases are detected, stride time, cadence, average velocity and step length gait parameters can be calculated. These parameters were calculated as in (Sabatini et al. 2005; Henriksen et al. 2004).

2.3.2 Assessment of the Fall Risk

The evaluation of the risk of falling of the subject is reached by an accelerometer attached to the trunk. All the processing applied in these signals is adapted from (Doheny et al. 2012). A band-pass filter of fifth order between 0.1-10Hz filters the signals, to restrict the signal. Then, to obtain the displacements of the subject's COM the acceleration signals are double integrated, using a trapezoidal method. The error associated to the integration (low frequency drift) is reduced by subtracting the mean of the acceleration signals before and after each integration, and then implementing a second-order polynomial fit and a high-pass filter of fifth order to 0.1Hz.

These signals enable to determine the Root Mean Square (RMS) for AP and ML directions, sway range AP and Ml and horizontal displacement of the COM (*Dhor*), given by,

$$Dhor = \sqrt{dML^2 + dAP^2} \quad (10)$$

where *dML* and *dAP* are obtained after both integrations and correspond to the displacements in AP and ML directions, respectively. Then, it is calculated the horizontal displacement of the COM. These parameters enable to assess the risk of falling of the patient.

2.3.3 Statistical Analysis

For each parameter the mean and standard deviation was calculated. Then, a Student's t-test was performed to compare the results obtained with the crutches and standard walker with the RFS. The level of significance was set to $p<0.05$.

3 RESULTS

Figure 4 show portions of the signals of one of the patients with the crutches, standard walker and RFS, acquired with the accelerometer attached to the ankle. As mentioned before, it was used one FSR to validate the detection of the gait events. The FSR detects 60ms and 30ms earlier the HS and TO events, respectively. In the three graphs, it is indicated, by different markers, both the instants of HS and TO detected by the accelerometer and by the FSR. It is also identified some of the gait parameters determined in the study. Figure 5 a and b are a portion of the AP and ML accelerations of the trunk, respectively, with each AD, acquired by the accelerometer placed at the sacrum. It is discriminated the AP and ML accelerations. Table 1 presents the mean and standard deviation of each parameter for 7 patients for the three ADs. As one can see in Table 1, the values of p-values less than 0.05 were obtained for the same parameters among the different ADs (stride and stance time, velocity, cadence and sway range ML). Relatively to the values, the crutches provide the higher stride, stance, swing time and step length. The RFS has the lower values for these parameters, except for the velocity and cadence. Considering the values acquired by (Martins et al. 2013) with a similar RFS, but with laser sensor, for the same diagnosis, one can see that they have obtained similar values.

In terms of the parameters obtained by the accelerometer at the trunk, all of them were greater for the crutches and smaller for the RFS, except for the RMS AP.

4 DISCUSSION

The goal addressed in this work is to detect differences between the ADs for the assisted gait. Hence, the gait parameters are studied in order to verify which are most affected in patients with KOA, relatively to gait events and fall risk. Furthermore, it is intended to verify and validate if the data extracted from the accelerometer is able to detect gait events in assisted-gait. As it can be seen in figure 4 for each gait cycle it is possible to observe two peaks (HS and TO).

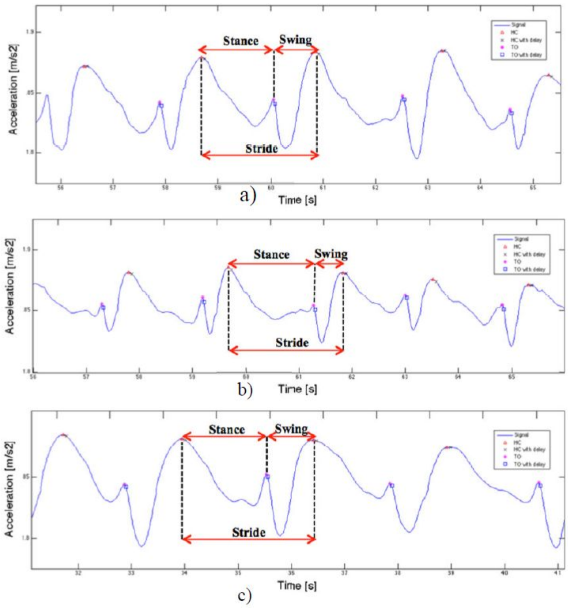

Figure 4: Portion of the signals of the ankle of a patient walking with a) crutches b) standard walker and c) RFS. The x-axis and y-axis correspond to the time and acceleration, respectively. The red triangle and the black cross correspond to HS and HS with delay, respectively. The rose star and blue square are relative to the TO and TO with delay. These graphics are in accordance to (Lee et al. 2010).

For the assisted gait with crutches, standard walker and RFS, the moment of HS occur at t=58.5s, t=59.7s and t=34.0s and TO at t=60.1s, t=61.3s and t=35.5s, respectively. This means that for each gait cycle and for each device, it can be detected the time that the foot contacts the ground and the time that it leaves the ground, respectively. Hence, with the identification of these two events, it is possible to determine the desired gait parameters. The recorded signals for walking trials are very similar to those presented in the literature for free walking (Lee et al. 2010). Therefore, it was reasonable to take into account these signals for further analysis.

Thus, it can be concluded that accelerometers can effectively be used in the gait analysis of assisted gait since they verify a relationship between the measured acceleration signal and gait events.

By observation of Table 1, comparing the stance

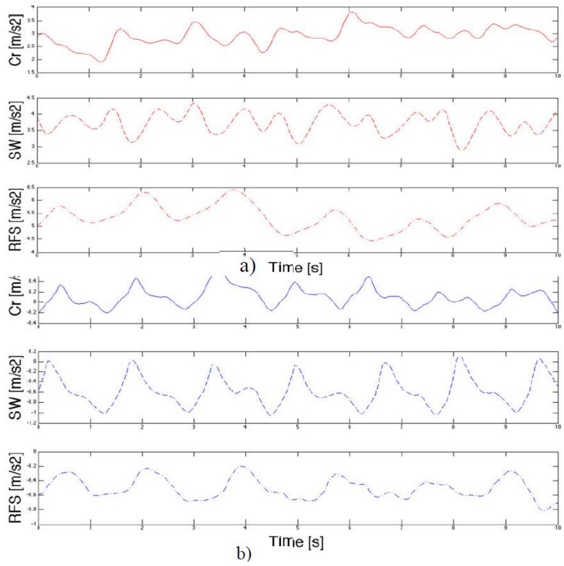

Figure 5: Portion of the signals a) AP and b) ML of the trunk for one patient. The first, second and third for each graph, correspond to crutches, standard walker and RFS, respectevely.

Table 1: Devices used and calculated parameters. The mean and the standard deviation for the parameters for assisted gait with the three ADs are listed. The values between brackets correspond to the percentage of the gait cycle. D:Device, P:Parameters, Cr:Crutches, p:p-value, SW:Standard Walker, ST:Stride Time, SgT:Swing Time, StT:Stance Time, C:Cadence, v:Velocity, SL:Step Length, DC:Displacement COM, R AP:RMS AP, R ML:RMS ML, SR AP:Sway Range AP, SR ML:Sway.

D	P	Cr	p	SW	p	RFS
Acc.(Ankle)	ST (s)	4.39 ±1.99	0.03	3.73 ±1.12	0.01	2.65 ±0.66
	SgT (s)	1.42 ±0.53 (32.35%)	0.22	1.21 ±0.57 (32.44%)	0.65	1.12 ±0.31 (42.26%)
	StT (s)	2.93 ±1.56 (67.65%)	0.03	2.51 ±0.59 (67.56%)	0.005	1.47 ±0.67 (57.74%)
Camera	C (step/ min)	25.76 ±13.86	0.008	30.59 ±10.16	0.002	43.35 ±14.12
	v (m/s)	0.11 ±0.05	0.02	0.09 ±0.04	0.004	0.20 ±0.03
	SL (m)	0.31 ±0.10	0.26	0.29 ±0.18	0.96	0.29 ±0.09
Acc.(Trunk)	DC (m)	1.17 ±0.49	0.13	0.69 ±0.19	0.57	0.52 ±0.28
	R AP (m/s^2)	0.69 ±0.24	0.49	0.49 ±0.08	0.56	0.61 ±0.29
	R ML (m/s^2)	0.48 ±0.34	0.42	0.40 ±0.10	0.16	0.32 ±0.06
	SR AP (m)	1.95 ±1.25	0.21	1.14 ±0.30	0.62	0.89 ±0.56
	SR ML (m)	1.08±0.13	0.01	0.77±0.18	0.007	0.52 ±0.17

and swing percentages for the crutches, standard walker and RFS, one can see that the stance phase is 67.65%, 67.56% and 57.74% and the swing phase is 32.35%, 32.44% and 42.26%, respectively. Considering that normal free gait is characterized by having 60% of stance phase and 40% of swing phase (Vaughan et al. 1999), the results obtained with RFS are the more approximated to these normal values. The stance (58.06%) and swing (41.92%) phases percentage are similar to the ones obtained by Martins et al. (2013) and they justified these values by the fact that these patients are better supported by the RFS, and they feel less pain when loading the affected joint, allowing to perform a more natural gait. The greater swing phase percentage with the RFS, relatively to the others ADs, could be explained by the existence of the forearm supports, which provide a greater support. Relatively to the stance phase percentage, it was verified a decrease with the RFS, comparatively to the others ADs. Since impaired gait and/or fear of falling usually results in an increase of stance time (Kloos et al. 2012), this result shows that the RFS offered excellent support and stability for the user, by increasing his sense of security relatively to the others ADs. The crutches produced the greater stride time and, consequently, greater stance and swing time, relatively with the others ADs, which is good, because it means that the patient spent more time with the leg, that has the knee injured, in the ground – stance phase. On the other side, the RFS the lower values for the stride, stance and swing time, because of the continuous movement of the subject with this device. The value of the velocity for assisted gait for the patients is lower than for the healthy (Martins et al. 2013). Considering our values, the standard walker shows the lower value for velocity. This may be explained because, to walk with this device, the patient has to stop, lift the AD and move forward, performing an unnatural gait. On the other side, the RFS has the higher values for velocity and cadence. So, this device is the nearest to the healthy and can be explained by the continuous movement of the subject with the RFS.

Finally, the step length is almost identical for the ADs, so devices preserve this feature. The little increase of the step length for the crutches may be reached by the fact that these patients have already walked with crutches before this study.

In terms of the evaluation of the fall risk, in (Kloos et al. 2012) it is mentioned that the variability is an indicator of fall risk, which means that the increase of variability increases the risk of fall. Thus, the variability of the stride, stance and swing time (Table 1) was analysed in this study for further fall risk analysis. As one can see, the standard deviation is lower for the gait with the RFS than the others ADs. Therefore, it can be verified that the RFS provides a greater stability for the patient. Relatively to the parameters obtained by the accelerometer at the trunk, it is known that the greater they are, the greater is the risk and trend of fall (Doheny et al. 2012). In Table 1, it is shown that the crutches present the higher values, so these devices are of higher risk to the user and the RFS produced the lower values, except for RMS AP. Considering the AP signal, it corresponds to the forward and backward movement of the trunk. One can see in Figure 5 a) that for the three devices the signal is positive, meaning that the trunk is leaning forward when the patient walks. However, the signal is much higher for the RFS than the others ADs. This happens because the RFS has to be pushed and by observation of the authors, users had to lean forward while pushing this device. This factor can be due to an incorrect walker height adjustment. Thus, the AP signal can be an important indicative for posture correction as well as walker height adjustment. Therefore, further studies will be conducted to evaluate this potential clinical indicator. In terms of ML signal in Figure 5 b), the RFS presented the lower values, relatively to the other ADs. The ML movement is a little attenuated by the RFS and this happens because the user is supported by the forearm support of RFS, preventing the trunk oscillation in this direction.

Finally, relatively to the variability of the signals of the trunk, it is shown in Table 1, that crutches present a higher standard deviation for all of the parameters, except for the sway range ML and RMS AP. The first is higher for the standard walker. The latter is higher for the RFS as expected, because of the leaning of the trunk to the front.

To conclude, the authors see the RFS as the best device for these patients since it provides higher stability to the users, less risk of fall, a more natural gait and a continuous movement.

5 CONCLUSION

This work used accelerometers located at the injured leg's ankle and trunk to verify and validate the association between the accelerations signals and the gait events, detect gait parameters and assess the fall risk in assisted-gait with crutches, standard walker and RFs. Further, it was possible to determine efficiently all the proposed gait parameters in all devices with patients diagnosed with KOA. Additionnaly, it can be verified that the RFS provides a greater stability, reducing the risk of fall and inducing a more natural gai performance.

ACKNOWLEDGEMENTS

This work has been supported by FCT – Fundação para a Ciência e Tecnologia in the scope of the project: PEst-OE/EEI/UI0319/2014.

REFERENCES

Debi, R. et al., 2011. Correlation between single limb support phase and self-evaluation questionnaires in knee osteoarthritis populations. *Disability and rehabilitation*, 33(13-14), pp.1103–9.

Doheny, E. P. et al., 2012. Displacement of centre of mass during quiet standing assessed using accelerometry in older fallers and non-fallers. *Annual International Conference of the IEEE Engineering in Medicine and Biology Society*. 2012, pp.3300–3.

Elbaz, A. et al., 2012. Can single limb support objectively assess the functional severity of knee osteoarthritis? *The Knee*, 19(1), pp.32–5.

Henriksen, M. et al., 2004. Test – retest reliability of trunk accelerometric gait analysis. *Gait & Posture*, 19, pp.288–297.

Kaufman, K. R. et al., 2001. Gait characteristics of patients with knee osteoarthritis. *Journal of biomechanics*, 34(7), pp.907–15.

Kloos, A. et al., 2012. The impact of different types of assistive devices on gait measures and safety in Huntington's disease. *PloS one*, 7(2), p.e30903.

Lee, J.-A. et al., 2010. Portable activity monitoring system for temporal parameters of gait cycles. *Journal of medical systems*, 34(5), pp.959–66.

Martins, M. et al., 2013. Assessment of walker-assisted human interaction from LRF and wearable wireless inertial sensors. In *Internation Congress on Neurotechnology, Electronics and Informatics. Neurotechnix*. pp. 1–8.

Mathie, M. J. et al., 2004. Accelerometry: providing an integrated, practical method for long-term, ambulatory monitoring of human movement. *Physiological Measurement*, 25(2), pp.R1–R20..

Sabatini, A. M. et al., 2005. Assessment of Walking Features From Foot Inertial Sensing. *IEEE transactions on bio medical engineering*, 52(3), pp.486–494.

Vaughan, C., Davis, B. & Connor, J. C. O., 1999. *Dynamics of Human Gait* 2nd ed. C. Vaughan, ed., Cape Town, South Africa: Kiboho Publishers.

Watanabe, T. et al., 2011. A preliminary test of measurement of joint angles and stride length with wireless inertial sensors for wearable gait evaluation system. *Computational intelligence and neuroscience*, 2011, p.975193.