Gait & Posture 37 (2013) 223–228

Contents lists available at SciVerse ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Intention detection of gait initiation using EMG and kinematic data

E.C. Wentink [a,*], S.I. Beijen [a], H.J. Hermens [a,b], J.S. Rietman [b,c], P.H. Veltink [a]

[a] Biomedical Signals and Systems group, MIRA Institute for Biomedical Technology and Technical Medicine, University of Twente, The Netherlands
[b] Roessingh Research and Development, The Netherlands
[c] Biomedical Engineering, MIRA Institute for Biomedical Technology and Technical Medicine, University of Twente, The Netherlands

ARTICLE INFO

Article history:
Received 9 August 2011
Received in revised form 14 March 2012
Accepted 13 July 2012

Keywords:
Electromyography
Transfemoral
Intention detection
Gait initiation

ABSTRACT

Gait initiation in transfemoral amputees (TFA) is different from non-amputees.

This is mainly caused by the lack of stability and push-off from the prosthetic leg. Adding control and artificial push-off to the prosthesis may therefore be beneficial to TFA.

In this study the feasibility of real-time intention detection of gait initiation was determined by mimicking the TFA situation in non-amputees. EMG and inertial sensor data was measured in 10 non-amputees. Only data available in TFA was used to determine if gait initiation can be predicted in time to control a transfemoral prosthesis to generate push-off and stability. Toe-off and heel-strike of the leading limb are important parameters to be detected, to control a prosthesis and to time push-off.

The results show that toe-off and heel-strike of the leading limb can be detected using EMG and kinematic data in non-amputees 130–260 ms in advance. This leaves enough time to control a prosthesis. Based on these results we hypothesize that similar results can be found in TFA, allowing for adequate control of a prosthesis during gait initiation.

1. Introduction

Gait initiation in transfemoral amputees (TFA) is different from non-amputees [1,2]. In non-amputees it consists of two phases. First, preparations are made for the step execution [1,3,4]. During this phase postural adjustments are made, the center of pressure moves towards the leading limb and the body is tilted forward. Subsequently the center of pressure moves towards the trailing limb and the body is tilted further forward. The hip and knee of the leading limb start to flex and the ankle starts to dorsiflex to prepare for toe-off, which is the end of the first phase. In the second phase the step is executed. It starts at toe-off of the leading limb and the body is tilted further forward. Muscles in the trailing limb stabilize the body, during swing of the leading limb, and generate push-off. The execution phase ends at heel-strike of the leading limb [1–5].

In TFA these two phases are similar, but the duration differs depending on which leg is leading, the prosthetic leg or the sound leg. It appears that TFA have the tendency to stand on their sound leg for as long as possible and load the prosthesis as short as possible [1,2,5].

Artificial push-off of a transfemoral prosthesis during gait initiation may be beneficial, to allow a more natural process and reduce effort needed from the sound leg [2]. However, gait initiation must be predicted, because timing of push-off is very important. Push-off in gait is described as the part of the gait cycle which begins at onset of ankle plantar flexion and ends at toe-off [6]. Starting push-off too early will propel the amputee backwards. Starting too late will dissipate push-off or even cause a stumble. To provide control inputs for supported prosthetic gait initiation, the beginning and end of the execution phase, toe-off and heel-strike of the leading limb respectively, need to be detected for both leading limb conditions. If in amputees the prosthetic leg is leading, the prosthetic knee should flex at toe-off and be ready to take the load at heel-strike. When the prosthetic leg is trailing, the prosthesis should provide push-off [3,7].

For the detection of gait initiation several sensors may be used like gyroscopes and accelerometers, but also electromyography (EMG) from the remaining muscles. EMG of gait initiation in non-amputees was measured in several studies but primarily at the lower leg [3,8,9]. EMG activity in amputees during gait has been measured in a few studies and is comparable to that of non-amputees [10–12]. EMG during gait initiation in TFA has not been studied previously.

Inertial sensors have frequently been used to assess gait. However, few studies were found that used inertial sensors to assess gait initiation [13].

Most studies used a combination of an optical position measurement system and force plates [1–3,5,8]. The authors found no studies on real-time intention detection of gait initiation in (non-)amputees. We therefore studied gait initiation detection

* Corresponding author at: Biomedical Signals and Systems group, MIRA Institute for Biomedical Technology and Technical Medicine, University of Twente, P.O. Box 217, 7500 AE Enschede, The Netherlands. Tel.: +31 534892766.
E-mail address: e.c.wentink@utwente.nl (E.C. Wentink).

0966-6362 © 2012 Elsevier B.V. Open access under the Elsevier OA license.
http://dx.doi.org/10.1016/j.gaitpost.2012.07.013

in non-amputees, before advancing to TFA, but mimicking the TFA situation. We used data which can be measured in TFA, i.e. upper leg muscle activity and inertial sensors. Therefore the data can be used for upper leg prosthesis, lower leg prosthesis or even orthosis. In these applications the need for stability and control is desired in order to improve gait initiation and gait [1,2,13–15].

The goal of this study is to determine if gait initiation can be detected from EMG of the upper leg muscles and/or inertial sensors. Detection should be sufficiently early to eventually support gait initiation in transfemoral prostheses users. The current study was performed in non-amputees.

2. Methods

2.1. Participants

Ten healthy volunteers participated in the study, none of them had a history of lower limb injuries, neurodegenerative diseases or any skin conditions. The experiments were approved by the local Ethics Committee and an informed consent was obtained before the experiments.

2.2. Measurements

Kinematic data was measured (100 Hz) using 2 inertial sensors from Xsens (Enschede, Netherlands), with 3D accelerometers, gyroscopes and magnetometers. Electrodes were placed according to the SENIAM standards [16]. On each muscle two self adhesive electrodes (Kendal, H93SG, Tyco healthcare, Germany) were placed as closely together as possible. EMG measurements were performed with a 16 bipolar channel Porti-system from TMSi (Oldenzaal, Netherlands) at a sample frequency of 2048 Hz, no pre-filtering was applied.

Nine muscles were measured, due to a limited number of available EMG channels and to mimic the prosthetic situation. The upper leg muscles and inertial sensors were placed on one leg, which is the "simulated prosthetic leg". The measured upper leg muscles are: the m. gluteus maximus (GMa), m. gluteus medius (GMe), m. tensor fasciae latae (TFL), m. rectus femorus (RF), m. vastus lateralis (VL), m. biceps femoris (BF). In five subjects one extra muscle, the m. erector spinea (ES), was measured.

At the lower leg on the contralateral side the m. tibialis anterior (TA), m. gastrocnemius medialis (GaM), m. soleus (Sol) were measured, for reference purposes. This is the "simulated sound leg". In five subjects the simulated sound leg was the dominant leg and in the other five it was the non-dominant leg.

Footswitches, placed mid-heel and at the first metatarsal head of each foot, gave information about heel-strike and toe-off. Subjects wore their own low-heeled shoes. Fig. 1 illustrates the placement of the inertial sensors and EMG electrodes. To synchronize EMG, footswitches and inertial sensors a synchronization pulse was given at the start and end of each measurement which was visible in all data sets.

2.3. Procedures

For the gait initiation experiments the subjects were required to stand upright with their weight equally distributed on both feet, the initial posture. Data recording was started. After 5 s in the initial posture the subjects were asked to press the synchronization button (sync) and start walking. After five paces they were asked to stop, turn around, return to the initial posture, wait 2–3 s, press the sync and walk back. This was repeated four times within each measurement. Two measurements were performed for each leading limb condition, 16 trials per condition.

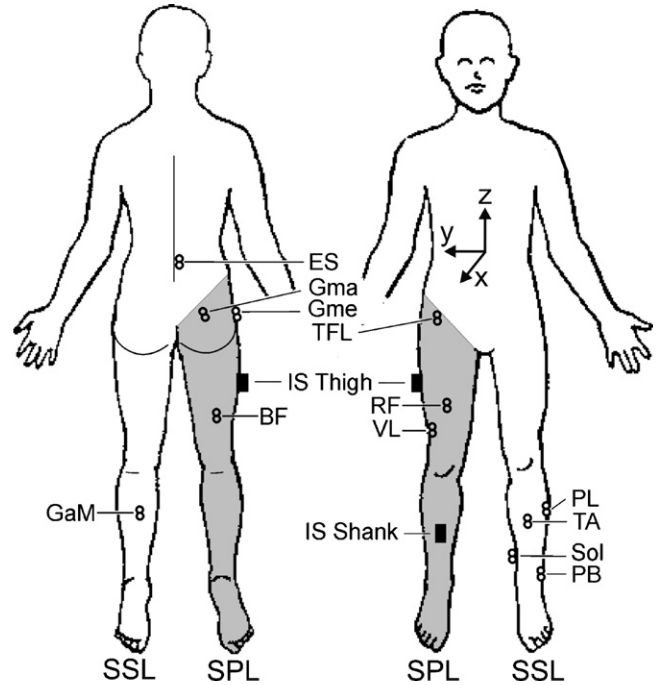


Fig. 1. Placement of the inertial sensors (IS) and EMG electrodes on the body. One leg simulates the prosthetic leg (SPL, in gray). At this leg all the upper leg muscles were measured and the inertial sensors were placed at the upper and lower leg. At the other leg, the simulated sound leg (SSL), only the lower leg muscles were measured for reference.

In addition a postural sway measurement was performed, a forward and backward swaying motion, without falling forward or backward. This was used to calculate the thresholds for the inertial sensors for gait initiation detection.

2.3.1. Sensor to body calibration

The inertial sensors express their data in the sensor coordinate system ($\vec{s}$). Two calibration exercises were performed to convert this data to the body coordinate system ($\vec{f}$), using the rotation matrix (R_{fs}) ($\vec{f} = R_{fs}\vec{s}$). In short the calibration of the lower leg was as follows. The subject stood upright, whereby the body z-axis equals the gravity vector which can be described in the sensor coordinate system. Subsequently the subjects flexed the knee five times to about 90°, where the knee is the body y-axis, allowing this axis to be defined in the sensor coordinate system. The x-axis is subsequently obtained by a cross product of the y and z axes. A similar procedure was followed for the upper leg sensor, using the squat as calibration exercise [17,18]. This data was subsequently low-pass filtered at 10 Hz. Finally the axis with largest amplitude was used for further analysis. For the angular velocity this was the body y-axis, for the acceleration it was the body z-axis (see Fig. 1).

2.4. Data analysis

EMG data analysis was performed in two parts. First the linear envelopes of the ensemble averages were calculated, to determine which muscles show a clear change in activity before toe-off or heel-strike of the leading limb. Secondly, from the selected muscles the onset or offset timings were determined.

The EMG data was first high-pass filtered using a second order Butterworth filter with a cut-off frequency of 20 Hz [19]. To calculate the linear envelopes the data was subsequently rectified and low-pass filtered with a second order Butterworth filter at 9 Hz [19]. To calculate the timings, the high-pass filtered data was low-pass filtered at 500 Hz [19].

Onsets of the selected muscles were calculated using a log-likelihood-ratio test (AGLR), as described by Staude [20,21]. This algorithm is suitable for real-time EMG onset or offset detection. The window-size used for the detection was 20 ms, the threshold of the algorithm for on-off detection was set at 20 [20–22].

The different phases of gait initiation were determined using the footswitch data. Push-off takes place between heel-off of the trailing limb, which is the onset of plantar flexion, and toe-off of the trailing limb [6]. For both leading and trailing limb heel-off, toe-off and heel-strike were determined. The data of each measurement was subsequently separated into trials and the trials were aligned at heel-strike of the leading limb. From the aligned trials the ensemble averages were calculated per subject.

The intra subject variability of the EMG trials was determined using the variance ratio for each subject and muscle for the preparation and the execution phase [8,23]. The variance ratio is the variance of the data between gait initiation cycles normalized to the total variance. The lower the score is, the higher the repeatability. Differences between the preparation and execution phase were analyzed using a paired t-test with a p-value of 0.05 and Bonferonni corrections [8].

Postural sway measurements were performed to determine thresholds for the kinematic data after calibration, to decide if the subject was performing postural sway or was initiating gait. If the data from the measurements exceeded the sway thresholds, then toe-off or heel-strike of the leading limb could be detected.

3. Results

3.1. Ensemble averages

Fig. 2 shows a typical example of the ensemble averages of the upper leg muscles and the inertial sensors, of one subject where the simulated prosthetic leg was leading (left) and where the simulated prosthetic leg was trailing (right). The arrows show the muscles and inertial sensors that have consistent activity changes before toe-off or heel-strike of the leading limb for all subjects.

The variance ratios of the different muscles in the preparation and execution phase are shown in Fig. 3(b). The execution phase is significantly better reproducible compared to the preparation phase in case the prosthetic limb was leading. The muscles that can be used for the detection of gait initiation show generally a lower variance ratio than the other muscles, but this is not significant. The standard deviations, the between subject variation, are large in some cases but do not differ between the different conditions.

3.2. Timings

Heel-strike of the leading limb was detected in all cases and was used as a reference measure for all timings, because toe-off of the leading limb was not detected in all trials. This was due to inaccurate or missing foot-switch data. Some trials were excluded, because subjects started walking with the wrong leg or no detection took place at all. The number of subjects and the percentage of trials included in the calculation of the timings are specified in Table 2.

Results for the timings of the footswitches can be found in Table 1. Total push-off time (SD) was 285 ms (75), starting 166 ms (66) before and ending 125 ms (38) after heel-strike of the leading limb. Table 2 shows the on and offset timings of the upper leg muscles before toe-off or heel-strike of the leading limb in all subjects. Table 2 also shows the detection of toe-off of the leading limb using inertial sensor data, which was only possible when the prosthetic leg was leading.

3.3. Simulated prosthetic leg leading

The TFL and the RF showed activity onset in the ensemble averages, which is confirmed by the average onset of 129–199 ms before toe-off of the leading limb. The VL and the BF showed onset of activity about 150 ms before heel-strike. Accelerometer and gyroscope data exceeded the sway thresholds 160–260 ms before toe-off. Heel-strike could not be predicted from the kinematic data, it could however be detected.

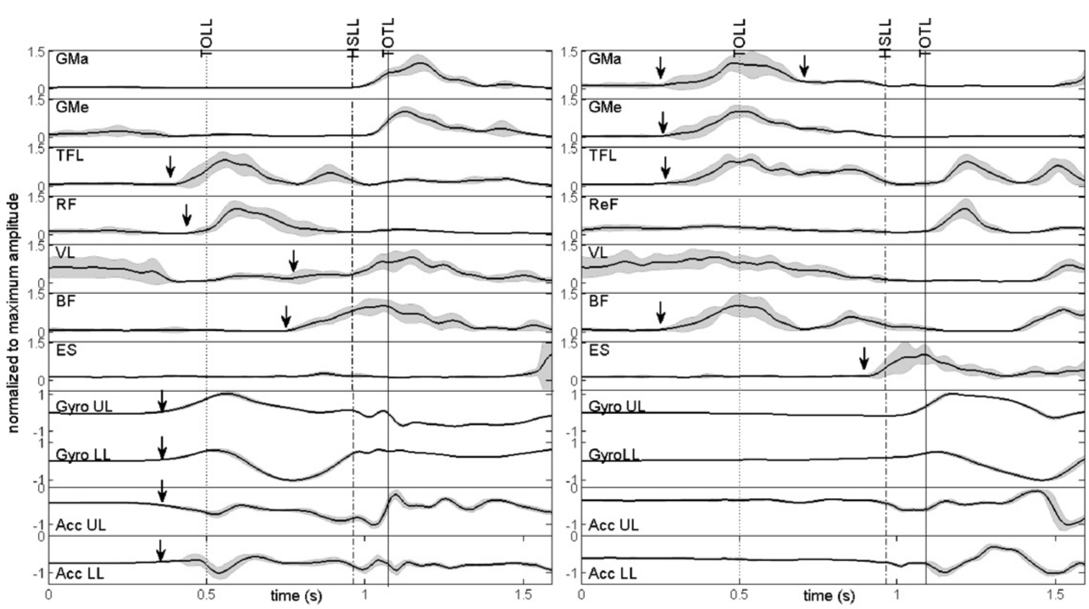
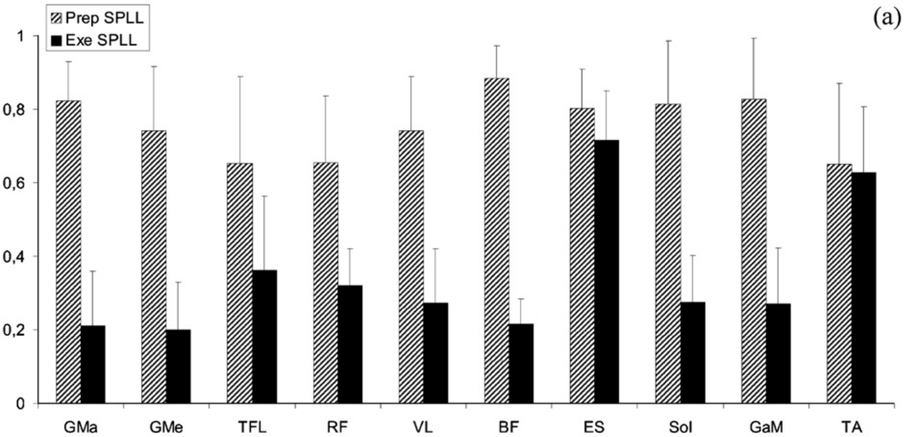


Fig. 2. Muscle activity of the upper leg muscles and inertial sensor data of the simulated prosthetic limb (SPL) during gait initiation. The ensemble average is taken over 16 trials of one typical subject. Thick black lines indicate the average activity and the gray surface indicates plus and minus one standard deviation. The vertical lines indicate the events: toe-off leading limb, heel-strike leading limb and toe-off trailing limb, respectively. On the left, where the SPL was leading, activity changes are seen before toe-off of the leading limb in the TFL, the RF and the inertial sensor data and in the VL and BF before heel-strike of the leading limb. On the right, the SPL was trailing activity changes are seen in the GMe, the GMe, the TFL and the BF before toe-off of the leading limb. The GMa and ES show activity changes before heel-strike of the leading limb.



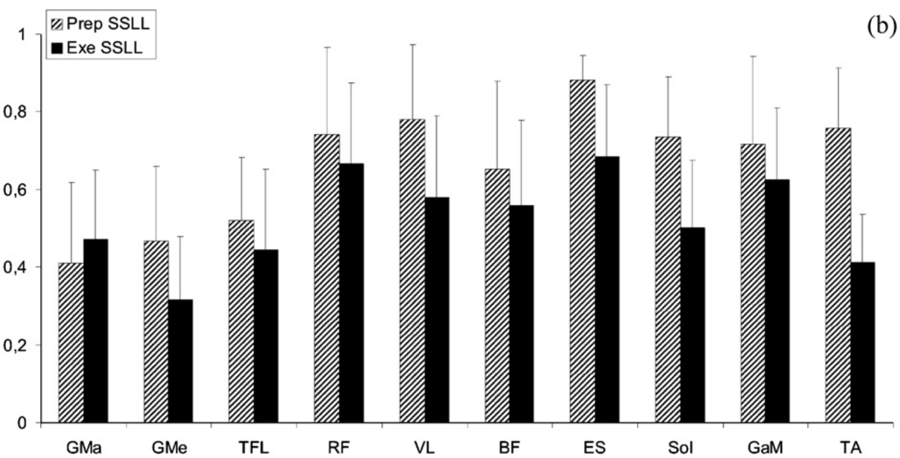


Fig. 3. Variance ratios of all muscles averaged over all subjects with one standard deviation. The lower the score is, the better the reproducibility. The preparation phase shows a significantly lower reproducibility than the execution phase when the simulated prosthetic limb was leading. (a) The reproducibility within subjects when the simulated prosthetic leg was leading and (b) when the simulated prosthetic leg was trailing.

3.4. Simulated prosthetic leg trailing

For this condition, the GMa, the GMe, the TFL and the BF could predict toe-off of the leading prosthetic limb 200–224 ms in advance. Heel-strike of the leading limb was detected for this condition by the GMa (offset) and the ES (onset) 163–199 ms in advance. Kinematic data could not be used to predict toe-off or heel-strike.

4. Discussion

The goal of the experiments was to determine if gait initiation can be predicted in non-amputees using data which would be available in prosthetic users on the prosthetic side, e.g. EMG and kinematic data. EMG of the upper leg muscles shows distinct patterns during gait initiation and was similar to that in other studies [3,8,9]. For both leading limb conditions EMG of the upper leg muscles showed activity changes 130–220 ms before toe-off and heel-strike. The RF and TFL can be used for the prediction of toe-off and the VL and BF for prediction of heel-strike of the leading (prosthetic) limb. The GMa, the GMe, the TFL and the BF can predict toe-off and the GMa and ES can predict heel-strike of the leading (sound) limb. Kinematic data could predict toe-off of the leading (prosthetic) limb, 158–260 ms in advance.

4.1. Simulated prosthetic leg leading

Previous studies showed that TFA have a tendency to start gait initiation with the prosthetic limb, because fewer adjustment strategies are needed to initiate gait with the prosthesis [1,2]. When the prosthesis is leading, the knee should flex at toe-off and extend at heel strike [3,7]. A short preparation and a long execution phase were seen if the prosthetic leg was leading compared to non-amputees [1]. But even if the preparation phase in TFA is half the

Table 1
Timing foot contacts.

	HOLL	TOLL	HOTL	HSLL	TOTL	HSTL	TOTL-HOTL	HOLL-TOLL
Time (ms)	−549	−462	−166	0	125	652	285	87
SD	49	49	66		38	18	75	61

Timings determined using the footswitches averaged over all trials of all subjects. TOLL, toe-off leading limb; HOLL, heel-off leading limb; HOTL, heel-off trailing limb; TOTL, toe-off trailing limb; HSLL, heel-strike leading limb; HSTL, heel-strike trailing limb. A minus sign refers to the event taking place before HSLL. Timings are averaged over all subjects.

Table 2
Timings of the upper leg muscles.

Leading limb	Muscle	On/off	TOLL (SD) in ms	# Sub	% Trials	HSLL (SD) in ms	# Sub	% Trials
SSL	GMa	On	−220 (97)	10	90	–	–	–
	GMe	On	−216 (49)	10	87	–	–	–
	TFL	On	−224 (62)	10	95	–	–	–
	BF	On	−200 (89)	10	81	–	–	–
	GMa	Off	–	–	–	−199 (70)	10	78
	ES	On	–	–	–	−163 (67)	5*	82
SPL	TFL	On	−129 (90)	10	82	–	–	–
	RF	On	−199 (108)	10	82	–	–	–
	VL	On	–	–	–	−145 (71)	9	88
	BF	On	–	–	–	−155 (45)	10	95
	Sensor		**TOLL (SD) in ms**	**# Sub**	**% Trials**			
	Acc	UL	−232 (34)	10	95			
	Acc	LL	−158 (90)	10	95			
	Gyro	UL	−260 (67)	10	95			
	Gyro	LL	−258 (34)	10	95			

A minus sign refers to the event taking place before the event. TOLL, toe-off leading limb; HSLL, heel-strike leading limb; SPL, simulated prosthetic leg; SSL, simulated sound leg; Acc, accelerometer data; Gyro, gyroscope data; UL, upper leg; LL, lower leg; #, the number of subjects included, %, the percentage of trials included, *, the ES was only measured in 5 subjects.

duration of that in non-amputees the current results suggest there is enough time to control the prosthesis. Due to the lower reproducibility of EMG in the preparation phase, the prediction of toe-off needs the inertial sensor data. In microprocessor controlled knees some of these inertial sensors are already build in. Heel-strike can be predicted using the EMG data.

4.2. Simulated prosthetic leg trailing

If in amputees the sound leg was leading the preparation phase was longer and the execution phase shorter compared to healthy individuals [1]. Due to the longer preparation there may be more time to detect toe-off of the leading limb in TFA compared to non-amputees. Timing of push-off, when the prosthetic limb is trailing, may need some consideration. To add push-off to prosthetic gait, heel-off and toe-off of the trailing limb need to be detected for correct timing [3,6,7]. However, if no ankle flexion takes place in the prosthesis heel-off and toe-off will occur almost at the same time. Detection of heel-strike and toe-off of the leading limb will be more useful. The results show that four muscles are able to predict toe-off of the leading (sound) limb with good reproducibility. However, only the GMa has a high reproducibility in heel-strike detection. Heel-strike of the leading limb may not need to be predicted (only detected) in this case, because push-off ends 125 ms after heel-strike.

4.3. Methodical considerations

Toe-off was not detected in all trials, the footswitches did not provide any information about the applied pressure. Furthermore, at the initial stance phase of gait initiation the weight of the subject shifts backwards a little which may unload the toe switches of the leading limb and therefore unloads the switches before actual toe-off.

The erector spinae was only measured in five subjects, during the experiments we found that the erector spinae may give valuable information on postural changes, therefore it was added later. Data of the ES may be used for detection of heel-strike if the prosthetic limb is trailing, but the variance ratios were among the highest. For the final application it is therefore not useful.

In previous studies, duration of activity of some muscles in TFA was found to be a little longer than in non-amputees [10–12]. This should not be a problem for gait initiation detection in TFA, as long as clear changes in muscle activity can be detected before toe-off and heel-strike of the leading limb. For offset detection of a muscle this may mean that less time is available prior to the event, but this was only relevant in the GMa if the simulated prosthetic leg was trailing. For longer stump lengths, amputation at the distal half of the upper leg, all the suggested muscles are likely to be available if myodesis of myoplasty has been performed. For short stump lengths, however, some of the suggested muscles may not be available any more [10].

Although all data was processed in such a way that onset detection can be performed real-time, there is need for a decision algorithm. To implement control into a prosthesis, similar results must be found in TFA and more activities should be analyzed.

4.4. Conclusions

Detection of gait initiation from EMG of the upper leg muscles and kinematic data in simulated amputee gait initiation was possible. Intention detection of gait initiation allows 130–260 ms for control of a prosthesis. However, further studies are needed to determine the possibilities to predict gait initiation in TFA.

Acknowledgments

This research is supported by the Dutch Technology Foundation STW, which is part of the Netherlands Organization for Scientific Research (NWO) and partly funded by the Ministry of Economic Affairs, Agriculture and Innovation, under grant no. 08003.

Conflicts of interest statement

Authors state that no conflicts of interest are present in the research.

References

[1] van Keeken H, Vrieling AH, Hof AL, Halbertsma JPK, Schoppen T, Postema K, et al. Controlling propulsive forces in gait initiation in transfemoral amputees. Journal of Biomechanical Engineering 2008;130(011002):1–9.
[2] Vrieling A, Keeken H, Schoppen T, Otten E, Halbertsma J, Hof A, et al. Gait initiation in lower limb amputees. Gait & Posture 2008;27:423–30.
[3] Mann R, Hagy J, White V, Liddell D. The initiation of gait. Journal of Bone and Joint Surgery 1979;61:232–9.
[4] Winter D. The biomechanics and motor control of human gait. Waterloo: University of Waterloo Press; 1987.
[5] Michel V, Chong R. The strategies to regulate and to modulate the propulsive forces during gait initiation in lower limb amputees. Experimental Brain Research 2004;158:356–65.
[6] Erdemir A, Piazza S. Rotational foot placement specifies the lever arm of the ground reaction force during the push-off phase of walking initiation. Gait & Posture 2002;15:212–9.

[7] Rose J, Gamble J. Human walking. Philadelphia: Lippincott Williams and Wilkins; 2006.
[8] Mickelborough J, van der Linden M, Tallis R, A.R. E. Muscle activity during gait initiation in normal elderly people. Gait & Posture 2004;19:50–7.
[9] Queralt A, Valls-Sole J, Castellote J. Speeding up gait initiation and gait-pattern with a startling stimulus. Gait & Posture 2010;31:185–90.
[10] Jaegers S, Arendzen J, de Jongh H. An electromyographic study of the hip muscles of transfemoral amputees in walking. Clinical Orthopaedics and Related Research 1996;328:119–28.
[11] Huang H, Kuiken TA, Lipschutz RD. A strategy for identification locomotion modes using surface electromyography. IEEE Transactions on Biomedical Engineering 2009;56(1):65–73.
[12] Ha K, Varol H, Goldfarb M. Volitional control of a prosthetic knee using surface electromyography. IEEE Transactions on Biomedical Engineering 2011;58 (1):144–51.
[13] Najafi B, Miller D, Jarett B, Wrobel J. Does footwear type impact the number of steps required to reach gait steady state?: an innovative look at the impact of foot orthoses on gait initiation. Gait & Posture 2010;32: 29–33.
[14] Tokuno C, Eng J. Gait initiation is dependent on the function of the paretic trailing limb in individuals with stroke. Gait & Posture 2006;24: 424–8.
[15] Hansen A, Miff S, Childress D, Gard S, Meier M. Net external energy of the biologic and prosthetic ankle during gait initiation. Gait & Posture 2010;31:13–7.
[16] Hermens H, Freriks B, Merletti R, Stegeman D, Blok J, Rau G, et al. European recommendations for surface electromyography. Enschede: Roessingh Research and Development; 1999.
[17] Luinge H, Veltink PH, Baten C. Ambulatory measurement of arm orientation. Journal of Biomechanics 2007;40:78–85.
[18] Zwartjes D, Heida C, van Vugt J, Geelen J, Veltink PH. Ambulatory monitoring of activities and motor symptoms in Parkinson's disease. IEEE Transactions on Biomedical Engineering 2010;57(11):2778–86.
[19] Merletti R, Parker P. Electromyography. Hoboken: Wiley Interscience; 2004.
[20] Staude G, Wolf W. Objective motor response onset detection in surface myoelectric signals. Medical Engineering & Physics 1999;21(6–7):449–67.
[21] Staude G, Flachenecker C, Daumer M, Wolf W. Onset detection in surface electromyographic signals: a systematic comparison of methods. In: EURASIP J Appl Sig Proc 2001, vol. 1. 2001. p. 67–81.
[22] Roetenberg D, Buurke J, Veltink P, Forner Cordero A, Hermens H. Surface electromyography analysis for variable gait. Gait & Posture 2003;18(2):109–17.
[23] Hershler C, Morris M. An optimality criterion for processing electromyography (emg) signals relating to human locomotion. IEEE Transactions on Biomedical Engineering 1978;25(5):413–20.