Estimation of the walking speed of individuals with transfemoral amputation from a single prosthetic shank-mounted IMU

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Abstract

Microprocessor prosthetic knees, able to restore the gait of people with transfemoral amputation, are now often equipped with sensors embedded in the prosthetic shank, which could be used to assess some gait characteristics during real-life activities. In particular, an estimation of the walking speed during the locomotion of those subjects would be a relevant indicator of the performance. However, if methods have already been proposed in the literature to compute this walking speed, none are directly usable in this context and with this population. For these reasons, the current study proposed to estimate the instantaneous walking speed with a shank-embedded Inertial Measurement Units based on a biomechanical model of the prosthetic lower limb. Averaged walking speed estimation has been quantified for nine individuals with transfemoral amputation walking on a treadmill at different speeds and slopes when wearing an instrumented knee ankle prosthesis. Experimental results demonstrated the ability of the model to estimate the walking speed with an accuracy of 9% (normalized root mean squared errors over all the patients), which is consistent with previous reported walking speed estimation errors. In addition, as the walking speed estimation is instantaneous, the proposed method can provide the estimation by the end of the stance phase, which is an originality compared to other methods based on step length estimation. The present method is relevant for the estimation of walking speed during real-life activities of above-knee amputees opening the way to direct activity monitoring from the prosthesis.

Keywords

Walking speed, ambulatory system, gait analysis, Inertial Measurement Unit; transfemoral amputee, center of mass

Introduction

During the last two decades, microprocessor prosthetic knees, able to control the stance and the swing phases of gait, have increased the safety and quality of life of individuals with transfemoral amputation.[1–3] However, to quantify prosthetic fitting objective performance requires the assessment of characteristics of the gait during real-life activities. Recently, Microelectromechanical systems technology has become affordable to develop wearable sensors that could monitor such real-life activities. Several information can be extracted from those sensors such as Inertial Measurement Units (IMU) going from segment angle estimation,[4–7] gait parameters estimation[8–10] to activity classification.[11–14] For people with lower limb amputation, cadence, step counts and activity bout duration have been the most used as estimators of in real life conditions.[15–21] On the contrary, walking speed estimation (WSE) has been rarely performed in people with amputation because of the difficulty to assess spatial parameters from wearable sensors signals. Thus, some authors have sought to link the cadence to the walking speed.[15] However, cadence is not always correlated with walking speed, for instance, when walking on different inclined surfaces as the step length changes.[22]

Nowadays, more and more microprocessor prostheses embed IMU in the prosthetic shank, which could be used to estimate the walking speed in real life

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environment. In the literature, several methods have been developed and evaluated on different populations.

Some authors proposed to use kinematic models of the lower limbs to perform this estimation. For asymptomatic subjects, Aminian et al. proposed a stance and swing phase kinematic planar model using three gyroscopes on one thigh, and on both shanks to estimate the walking speed. Aminian et al.'s[23] method was later adapted by Salarian et al.[24] with only shanks instrumentation. The main drawback of this method for people with amputation is the necessity to instrument the sound limb, which jeopardizes the compliance to such protocol. Other studies have proposed estimation of walking speed for individuals with transfemoral amputation using only one IMU and a gait model. Miyazaki[25] instrumented the thigh with a gyroscope and proposed a single segment planar model of the swing phase. In the study of Lenzi et al.,[26] an IMU was placed on the prosthetic shank but the kinematic model used was not fully described.

Other methods have been proposed for asymptomatic subjects. For example, machine learning was used by Aminian et al.[27] to estimate walking speed from a single axis accelerometer on one heel and a 3D one on the waist. If the method was proved to be accurate for one person, the generalization to other individuals is not straight forward due to overfitting and sensor placement sensitivity.[28,29] Another way to quantify the walking speed consists in double integrating a segment point acceleration. Authors using this procedure, generally placed the IMU on the foot,[30,31] benefiting from the stationary state of the foot during the stance phase. However, prosthesis embedded IMU are most often placed in the shank. Li et al. proposed a method of double integration to quantify the walking speed of asymptomatic people from an IMU positioned on the shank. However, this method assumes that the shank angular velocity is almost null at midstance.[32,33] This hypothesis is not adapted for people with transfemoral amputation due to the lack of knee flexion during early to midstance.[34] Plus, double integration of the acceleration can also be prone to random drift computationally heavy to remove for real time application.[35] Still, these methods have not been tested on this population.

In this context, the present study proposes to use a planar gait model to quantify the averaged walking speed by estimating the instantaneous velocity of the center of mass (COM) during the gait cycle using only one IMU integrated in a prosthetic shank. To evaluate this method, nine individuals with transfemoral amputation walked on a treadmill at different speeds and inclinations with a Microprocessor controlled Knee Ankle Prosthesis (MKAP) prototype.[36] Embedded sensors in the MKAP were used to compute WSE at each cycle, which was compared to the treadmill speed.

Methods

Kinematic model

The WSE method presented here is based on a kinematic planar model. It consists of an inverse pendulum model representing the prosthetic lower limb during the first half of the gait cycle (i.e. from the prosthetic heel strike to the contralateral heel strike) and considering the absence of knee flexion during this phase in the population of individuals with transfemoral amputation equipped with single axis knee prosthesis.[34,37] The pelvis, the thigh, the shank and the foot are modeled by a single rigid body going from the COM to the foot–floor contact point. The prosthetic foot-ankle complex is considered to be kinematically equivalent to a rigid arc shape rolling without sliding during stance.[38,39] From this model, the instantaneous COM velocity ($\overrightarrow{V_{COM}}$) expressed in the global frame (R_0) with $\overrightarrow{X_0}$ horizontal and aligned with the floor and $\overrightarrow{Y_0}$ vertical and pointing upward can be inferred from only four parameters: the COM height L when standing, the leg angular velocity $\dot{\theta}$ with respect to R_0, its angle θ with regard to the

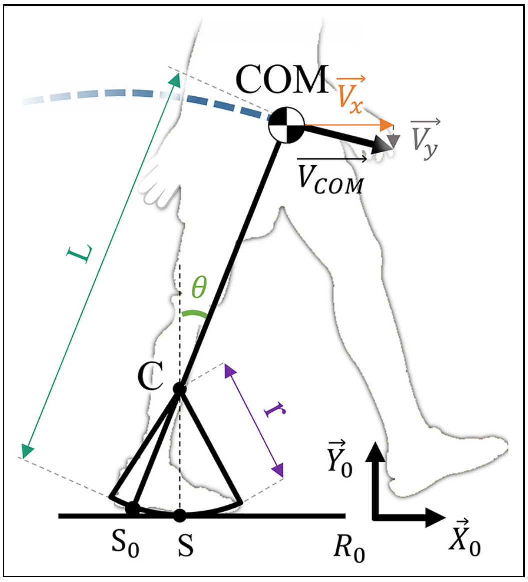

Figure 1. Stance phase gait model (S contact point between the foot and the floor, $S0$ contact point S when the lower limb is vertical, C: foot arc center; COM: center of mass; L: COM height; r: foot arc radius, $R0$ global frame ($\overrightarrow{Y_0}$ aligned with the gravity, $\overrightarrow{X_0}$ aligned with the walking direction), θ sagittal angle of the leg with respect to $R0$, $\overrightarrow{V_{COM}}$ = COM velocity, V_x = Projection of $\overrightarrow{V_{COM}}$ on $\overrightarrow{X_0}$, V_y = projection of $\overrightarrow{V_{COM}}$ on $\overrightarrow{Y_0}$).

gravity vector and the foot arc radius r, with C its center and S the foot–floor contact point (Figure 1)

$$\overrightarrow{V_{COM}} = \vec{V}(COM/R_0)_{R_0} = \vec{V}(C/R_0)_{R_0} + \vec{\Omega}(R_1/R_0)_{R_0} \times \overrightarrow{CCOM}_{R_0} = \dot{\theta} \begin{bmatrix} r + (L-r)\cdot\cos(\theta) \\ -(L-r)\cdot\sin(\theta) \\ 0 \end{bmatrix} \tag{1}$$

where $\times$ denotes the cross product. $\vec{V}(COM/R_0)_{R_0}$ is the speed of the COM relative to the global frame (R_0) expressed in the global frame (R_0), $\vec{V}(C/R_0)_{R_0}$ is the speed of the center of the foot arc expressed in R_0, $\vec{\Omega}(R_1/R_0)_{R_0}$ is the rotational speed of the Frame R_1 linked to the rigid body relative to R_0, and $\overrightarrow{CCOM}_{R_0}$ is the vector displacement from the point C to COM expressed in R_0. $\overrightarrow{V_{COM}}$ can be decomposed into V_x and V_y corresponding to the COM instantaneous velocity projection along the $\overrightarrow{X_0}$ axis and $\overrightarrow{Y_0}$ axis of the global frame R_0. Moreover, r and L can be estimated as proportions of the body height (BH), respectively, 19%,[40] and 58.8%[41] giving a simplified expression of the previous equation (2)

$$\vec{V}(COM/R_0)_{R_0} \cdot \overrightarrow{x_0} = V_x = \frac{BH \cdot \dot{\theta}}{100} \cdot (19 + 39.8 \cdot \cos(\theta)) \tag{2}$$

Equation (1) gives an instantaneous WSE during the stance phase on the affected side independent on the floor inclination as far as the model assumptions are verified (i.e. knee extended during stance). The average walking speed over the whole gait cycle is assumed to be equal to V_x (equation (2)) averaged over the first half of the gait cycle. To determine the gait cycle, heel strikes were detected with an ankle moment sensor considering a threshold of -5 Nm. $\dot{\theta}$ and θ were obtained thanks to the IMU embedded in the prosthetic shank using a complementary filter.[6]

Experimental protocol

Nine individuals with transfemoral amputation wearing the MKAP prototype participated in this study. Their characteristics are listed in Table 1. Different treadmill speed and slope conditions were tested by all participants depending on their capacity. All patients performed level walking at speeds ranging from 0.56 m/s to 1.39 m/s by 0.28 m/s increments apart one subject who didn't perform the 1.39 m/s condition. Three patients also walked at 1.67 m/s. Six of them also tested a 5.7° slope at their self-selected speed depending on their capacity. The values of walking speed are given in Table 1. For each speed and slope condition, 7–10 cycles were recorded after the steady state was established (more than 20 steps after the treadmill speed was changed). Between 1 and 5 min rest was allowed between each condition. The protocol was approved by the local ethics committee (RCB 2011-A00409-32), and all participants gave their consent.

WSE has been computed for each cycle. WSE error was computed against the treadmill speed. The treadmill speed can vary depending on weight acceptance, treadmill belt deformation, and motor servo loop, this variation has been estimated to be less than 4%.

Data analysis

For each treadmill walking trial, WSE was computed at each cycle. Estimation error for level and slope walking was calculated for each patient with mean error between WSE and the reference velocity $V_{treadmill}$, standard deviation error, root mean squared error (RMSE $= \sqrt{\sum_{n=1}^{N}(V_{treadmill\,n} - WSE_n)^2/N}$ with N the number of samples) as well as normalized RMSE (NRMSE $= (\sqrt{\sum_{n=1}^{N}(V_{treadmill\,n} - WSE_n)/V_{treadmill\,n})^2/N})$ were computed across all walking cycle for each slope condition. The averaged RMSE and NRMSE, mean error and standard deviation error over all patients were used to evaluate the intersubject reproducibility for each slope conditions. The agreement error was computed with the squared Pearson's correlation coefficient.

Results

Figure 2 presents WSE and WSE error against the treadmill speed considered as the reference.

The WSE method showed a good agreement with the reference speed ($R^2 = 0.93$) for all level walking conditions. Mean NRMSE, across all patients, for level

Table 1. Subjects characteristics and walking conditions conducted.

Patient	Height (m)	Mass (kg)	Level walking speed range (m/s)	Up slope walking speed (m/s) at 5.71°
S1	1.86	73	0.56–1.67	0.83, 1.39
S2	1.94	85	0.56–1.67	1.11
S3	1.90	81	0.56–1.39	n/a
S4	1.87	83	0.56–1.39	n/a
S5	1.93	131	0.56–1.67	0.83
S6	1.59	61	0.56–1.39	0.83
S7	1.86	117	0.56–1.39	0.83
S8	1.57	60	0.56–1.11	n/a
S9	1.75	79	0.56–1.39	0.83

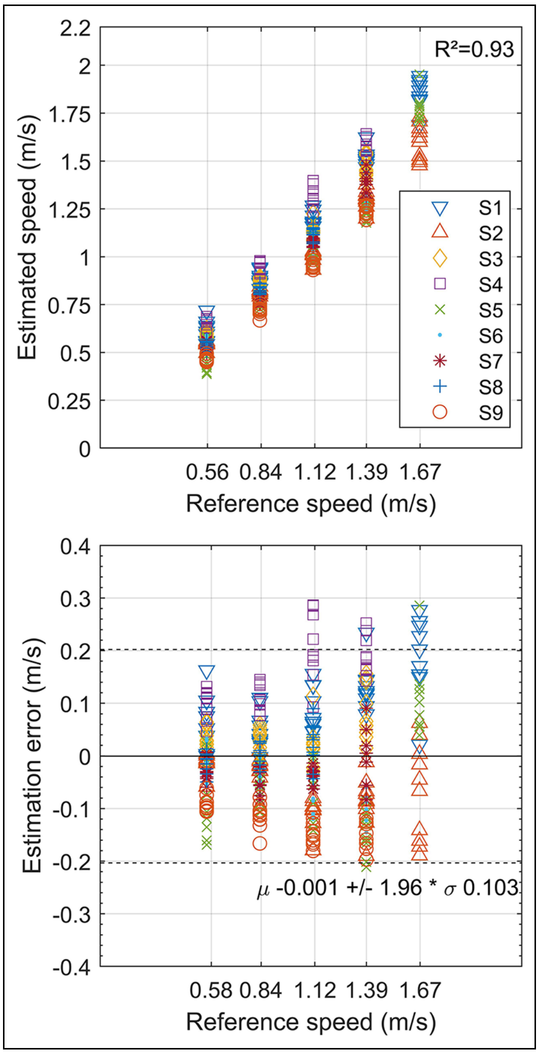

Figure 2. Top figure corresponds to WSE versus reference for each level walking speed conditions. Bottom figure corresponds to Bland—Altman plots for WSE. Each mark corresponds to a cycle WSE, each different mark corresponds to a different patient. R^2 corresponds to the overall correlation for all subjects cycles.

walking was 8.93% (Standard Deviation: 4.36%; range: 3.54%–16.53%). This result corresponded to a mean and standard deviation error of –0.002±0.088 m/s. The RMSE, NRMSE, mean and standard deviation errors at level walking can be found for each subject in Table 2. For the 5.7° slope condition, average NRMSE were 9.58% corresponding to 0.038 ± 0.050 m/s with a good agreement between the reference and the estimated walking speed (R^2 = 0.93). Errors for each patient for upslope walking conditions are reported in Table 3.

Table 2. Walking speed estimation errors for each patient for level walking conditions.

	RMSE (m/s)	NRMSE (%)	Mean error (m/s)	SD error (m/s)
S1	0.125	10.539	0.103	0.071
S2	0.088	6.820	–0.063	0.062
S3	0.059	5.918	0.043	0.041
S4	0.163	16.528	0.152	0.061
S5	0.109	11.234	–0.047	0.099
S6	0.071	6.663	–0.056	0.044
S7	0.046	5.178	–0.029	0.036
S8	0.031	3.537	0.003	0.031
S9	0.130	13.913	–0.126	0.033
Mean	0.091	8.926	–0.002	0.053
Standard deviation	0.044	4.363	0.088	0.022

RMSE: root mean squared error; NRMSE: normalized root mean squared error.

Table 3. Walking speed estimation errors for each patient for upslope (+5.7°) walking conditions.

	RMSE (m/s)	NRMSE (%)	Mean error (m/s)	SD error (m/s)
S1	0.118	11.091	0.110	0.044
S2	0.104	9.382	0.097	0.042
S5	0.161	19.378	0.124	0.108
S6	0.029	3.486	–0.027	0.012
S7	0.029	3.527	–0.023	0.020
S9	0.089	10.635	–0.053	0.075
Mean	0.088	9.583	0.038	0.050
Standard deviation	0.052	5.886	0.080	0.036

RMSE: root mean squared error; NRMSE: normalized root mean squared error.

Discussion

The current study presents a method to estimate the instantaneous walking speed based on a mechanical gait model using a single IMU located on the tibia and the BH. This walking speed was averaged during the stance phase to be compared with treadmill reference speed. The WSE error, obtained for individuals with transfemoral amputation, was inferior to 16.53% during level walking, which is consistent with the results of Lenzi et al.[26] (8% error, number of subjects not reported) and Miyazaki[25] (error inferior to 15%, number of subjects = 7). Mean WSE error is very close to the one obtained by Aminian et al.[23] with both lower limbs instrumented for twenty asymptomatic subjects or his modified single limb instrumentation method for ten asymptomatic subjects[24] (0.06 m/s error for both models). For their machine learning approach, Aminian et al.[27] only reported the maximum coefficient of variation on the distance estimated, making it difficult to compare the proposed method with theirs. Finally, the presented results are in the range of the double integration methods presented by Sabatini et al.[31] (0.05 m/s error) and Yang et al.[33] (4.2% error). Table 4 hereafter compares the WSE error reported in the literature according to the method used, the number of subjects and the population considered and the sensors and their placement.

Table 4. Walking speed estimation reported in the literature.

Author	*N* subject	Population	Sensors	Method	Results
Miyazaki[25]	7	TF	Thigh gyroscope	Kinematic model	Inferior to 15% relative error
Lenzi et al.[26]	?	TF	Shank IMU	Double integration	8%
Aminian et al.[27]	6	SA	Foot and waist IMU	Machine learning	Maximum coefficient of variation 6% on the total distance covered
Aminian et al.[23]	20	SA	Shank and thigh gyroscope on both legs	Kinematic model	RMSE 0.06 m/s (6.7%)
Sabatini et al.[31]	5	SA	Foot IMU	Double integration	RMSE 0.05 m/s
Yang et al.[33]	16	SA	Shank IMU	Double integration	RMSE 4.2%
Salarian et al.[24]	10	SA	Shank gyroscope on both legs	Kinematic model	0.04 ± 0.06 m/s
This study	9	TF	Shank IMU	Kinematic model	RMSE 0.09 m/s (9%) on level walking

TF: individuals with transfemoral amputation; SA: individuals without amputation; IMU: Inertial Measurement Units; RMSE: root mean squared error.

As concerns the instrumentation of the prosthesis, to our knowledge, previous methods, available in the literature and developed for asymptomatic subjects, almost all required sensors on other parts of the body or on the foot.[23,24,30] The foot placement of the sensor jeopardizes the use of a generic instrumented prosthetic component as the prosthetic foot must be adapted to each patient shoe size. Only Yang et al. proposed a method to compute WSE from a shank embedded IMU but it was not adapted for this population.[32,33] On the contrary, the model proposed in the present study was specifically designed for individuals with transfemoral amputation and revealed relevant for this specific population given the reported values of the estimation error.

An additional advantage of the method is to give an instantaneous WSE during the stance phase using only one IMU embedded in the prosthetic shank. In the literature, the estimation of the walking speed generally relied on the estimation of the step length, or foot displacement which could be obtained at the end of the gait cycle. Therefore, prosthetic behavior changes, based on this evaluation, cannot be effective before the next cycle.[23,33] On the contrary, the instantaneous WSE could be averaged at the end of the stance phase but could also be used to change the prosthesis behavior as soon as the swing phase starts.

Compared to the treadmill reference speed, the method underestimates the averaged walking speed. The uncertainty of the estimation of the COM height from the anthropomorphic model[41] directly affects the estimation of the COM velocity and could be evaluated to about $\pm 1\%$ of the BH.[42] In addition, as the kinematic model is planar, the pelvis transversal movements that have an impact on the COM position in the sagittal plane are neglected. Also, the method assumes the equality of the average COM speed on both half portions of the cycle. Asymmetry induced by gait deviations such as hip hiking, vaulting or pelvis transverse rotation[43] could also have an impact on this hypothesis. Results also showed that the estimation error was affected when walking in slope. The WSE error increased up to 9.58% for the mean NRMSE. A part of this error can be attributed to the forward velocity projection on the horizontal plane ($< 1\%$ in this study slope condition). Another part of this error could be due to the increase of the asymmetry in this situation.[44]

Finally, the WSE has been performed on a treadmill. Gait on treadmill is known to differ from overground walking.[45] However, the performance of the method should not be deteriorated overground as the assumptions made would not be disputed. Thus, even if the gait on treadmill is not completely representative of overground walking, we could expect similar results in the latest condition as the ones reported here.

To conclude, this study presents a method to estimate the walking speed for individuals with transfemoral amputation using a single shank embedded IMU. A mean 9% RMSE errors have been quantified with nine patients walking on a treadmill at a 0° slope. In a 5% slope, the RMSE errors slightly increased but remained acceptable for an estimation of the patient activity. The validation of the method should be completed by additional acquisitions in a real environment to show its relevance for the estimation of walking speed during real-life activities of above-knee amputees.

Declaration of conflicting interests

The author(s) declared the following potential conflicts of interest with respect to the research, authorship, and/or publication of this article: The corresponding author Boris Dauriac was an employee of Proteor, the company which developed the prosthesis used in this study. The other authors declare that there is no conflict of interest.

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