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Three-dimensional acceleration of the body center of mass in people with transfemoral amputation: Identification of a minimal body segment network

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A B S T R A C T

Background: The analysis of biomechanical parameters derived from the body center of mass (BCoM) 3D motion allows for the characterization of gait impairments in people with lower-limb amputation, assisting in their rehabilitation. In this context, magneto-inertial measurement units are promising as they allow to measure the motion of body segments, and therefore potentially of the BCoM, directly in the field. Finding a compromise between the accuracy of computed parameters and the number of required sensors is paramount to transfer this technology in clinical routine.

Research question: Is there a reduced subset of instrumented segments (BSN) allowing a reliable and accurate estimation of the 3D BCoM acceleration transfemoral amputees?

Methods: The contribution of each body segment to the BCoM acceleration was quantified in terms of weight and similarity in ten people with transfemoral amputation. First, body segments and BCoM accelerations were obtained using an optoelectronic system and a full-body inertial model. Based on these findings, different scenarios were explored where the use of one sensor at pelvis/trunk level and of different networks of segment-mounted sensors for the BCoM acceleration estimation was simulated and assessed against force plate-based reference acceleration.

Results: Trunk, pelvis and lower-limb segments are the main contributors to the BCoM acceleration in transfemoral amputees. The trunk and shanks BSN allows for an accurate estimation of the sagittal BCoM acceleration (Normalized RMSE $\leq 13.1\%$, Pearson's correlations $r \geq 0.86$), while five segments are necessary when the 3D BCoM acceleration is targeted (Normalized RMSE $\leq 13.2\%$, Pearson's correlations $r \geq 0.91$).

Significance: A network of three-to-five segments (trunk and lower limbs) allows for an accurate estimation of 2D and 3D BCoM accelerations. The use of a single pelvis- or trunk-mounted sensor does not seem advisable. Future studies should be performed to confirm these results where inertial sensor measured accelerations are considered.

1. Introduction

The study of biomechanical parameters derived from the body center of mass (BCoM) motion may reveal crucial information about gait impairment [1–3], especially in people with lower-limb amputation [3–5]. Indeed, 3D BCoM acceleration, velocity and displacement allow

to describe the kinematics of the body as a whole [6] and have been shown to provide insights on dynamic stability [7,8], gait efficiency [5, 9], and gait asymmetries [1,3] both in the able-bodied population and in lower-limb amputees. Although 3D BCoM motion is of particular interest to describe pathological gait, it is scarcely studied in clinical routine [3], partly due to the high cost and complexity of laboratory-based systems

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generally required for its investigation.

Recently, the use of magneto-inertial measurement units (MIMUs) has been proposed as an alternative to gold standard instrumentations for the estimation of BCoM-derived parameters [6,10,11]. MIMUs are indeed small, light, and low-cost wearable sensors, embedding orthogonally mounted 3D accelerometers, gyroscopes and magnetometers. The information provided by these sensors can be fused to estimate the orientation of the inertial frame defined by the MIMU case relative to an Earth-fixed frame [12]. Therefore, provided MIMUs are securely attached to their respective underlying segments with a known relative orientation [13], they can be used to estimate segmental orientation and motion, and ultimately the 3D motion of each segment's center of mass (SCoM) and of the whole body [6].

For the sake of simplicity, most wearable protocols developed for BCoM motion tracking involve a single sensor at pelvis level [11,14]. Yet, several works evidenced that the single-sensor paradigm tends to overestimate the 3D trajectory of the BCoM [2,14,15]. In particular, the mediolateral [15,16] and anteroposterior [14] components of BCoM position and acceleration were shown not to be accurately captured with this method, especially when participants exhibited an asymmetrical gait pattern or dynamical upper body motions, as it is the case in people with a lower-limb amputation [17]. Consequently, multi-segment methods, including 11–17 MIMUs, have been proposed [6,18,19].

If single-sensor approaches may not be accurate enough, especially for pathological gait, finding a balance between the number of MIMUs and the accuracy of BCoM-related parameters is essential to effectively allow the use of wearable protocols in clinical routine [11,15]. In this perspective, Shahabpoor and coworkers recently proposed a method to select a reduced number of MIMU-bearing segments (called an “optimal network of [sensors]”) for the estimation of 3D BCoM acceleration in able-bodied individuals using optoelectronic system data [10]. While the identified optimal segment configuration – including the trunk, pelvis and one thigh – achieved a satisfying accuracy in the vertical direction, only moderate accuracy was achieved in the mediolateral and anteroposterior directions (Normalized Root Mean Square Error = 7 % *versus* > 16 %) [10]. Although the adopted configuration may not suit transfemoral amputee gait due to differences in inertial parameters and overall gait pattern, the proposed method appears promising and could be adapted to this population in order to identify the location and minimal number of sensors required for an accurate monitoring of the BCoM acceleration of transfemoral amputee patients. On a longer-term perspective, such a methodology could be considered in-the-field to characterize clinically-relevant parameters derived from the BCoM acceleration such as the distribution of ground reaction forces between the prosthetic and sound leg [20] or the symmetry of external work production [21] which could benefit the rehabilitation by providing clinicians with quantitative data to monitor patients' progression or prescribe adapted prosthetic components.

The aim of the present study was therefore to identify potential optimal sensor networks, that is, subsets of segments to be instrumented with inertial sensors offering a good compromise between accuracy and number of sensors for the estimation of 3D BCoM acceleration in people with transfemoral amputation ambulating on level ground. First, segmental contributions to the BCoM acceleration were investigated using optoelectronic system data and a full body inertial model. This step allowed the identification of the most contributing segments to the BCoM acceleration. Based on these results, the BCoM acceleration was estimated from various combinations of the identified segments also called “body segment networks”.

2. Methods

2.1. Participants

The study was designed according to the Declaration of Helsinki and was granted ethical approval (Comité de Protection des Personnes CPP

NX06036). Ten people with traumatic transfemoral amputation (age: 41.5 ± 11.3 years; mass: 68.8 ± 15.2 kg; height: 1.73 ± 0.07 m; 8 males) gave written informed consent to participate in the study (Table 1). Inclusion criteria were adults with non-vascular transfemoral unilateral amputation, fitted with a definitive prosthesis and able to walk without any assistance. Participants walked with their usual passive microprocessor-controlled knee with an energy storing and return foot, the alignment of which was controlled by a prosthetist prior to data collection.

2.2. Measurement protocol

Each participant was equipped with a full-body marker set [8] (Fig. 1). An optoelectronic system (VICON, Oxford, UK, 200 Hz) recorded marker 3D positions while the participant kept a static standing posture and two photographs (front, profile) were taken. Following this static calibration, each participant walked at self-selected speed along an 8 m pathway, with 3 ground-embedded force plates (AMTI, MA, USA, 1000 Hz) in the middle. Only trials with three successive foot contacts on the force plates (i.e. a complete stride) were considered for further analysis.

2.3. Data processing

The markers positions were used to build a 15-segment hybrid inertial model, following [22] with the addition of hand segments [23], thus allowing to obtain body segmental inertial parameters (mass, inertial matrices and center of mass of each segment). Prosthetic limbs were represented by a concentrated mass estimated from the manufacturers' manuals similarly to [8] and located below the prosthetic knee joint and at the junction of the prosthetic foot and knee components. Markers and force plate data were filtered using a zero-phase fourth order Butterworth low-pass filter, with a cut-off frequency of 5 Hz. The acceleration of each segment center of mass (SCoM) and of the inertial-model-based BCoM were computed from differentiation of the marker-based signals, each differentiation step being preceded by the abovementioned low-pass filtering of the signals. Additionally, reference BCoM acceleration ($a_{BCoM,ref}$) was derived from ground reaction force time-series (*GRF*) (Eq. (1), with m_{body} , the participant's body mass and g the gravitational acceleration). All acceleration data were time-normalized to percent of the gait cycle, identified using a 20 N threshold on force plates data.

$$m_{body} a_{BCoM,ref} = GRF + m_{body} g \quad (1)$$

2.3.1. Segmental contributions

Segmental contributions to the BCoM accelerations were defined according to two criteria following [10]: the relative weight of SCoM accelerations in BCoM acceleration and the similarity between SCoM and BCoM acceleration patterns.

The weight of the contribution of each segment ($Contrib_{seg_i}$) in BCoM acceleration was defined as the SCoM acceleration (a_{SCoM_i}) weighted by the relative mass of the segment in the body (Eq. (2), m_{seg_i} for the i^{th} segment). Segmental contribution weights were normalized by peak-to-peak BCoM acceleration for each gait cycle.

$$Contrib_{seg_i} = \frac{m_{seg_i}}{m_{body}} a_{SCoM_i} \quad (2)$$

Regarding the similarity of SCoM accelerations, the average Pearson's correlation coefficients between each SCoM acceleration and the inertial-model-based BCoM acceleration were computed for each participant.

The contribution weights and correlations were averaged over all participants and used to identify the most contributing segments to BCoM acceleration.

Table 1

Participants' characteristics.

Participant	Gender	Age (years)	Height (m)	Mass (Kg)	BMI	Amputation delay (years)	Amputation level	Prosthetic knee	Prosthetic foot
TF1	M	58	1,8	68	21,9	31	TF	Mauch SNS	Variflex
TF2	M	48	1,8	64	19,7	1	TF	C-leg	Triton
TF3	M	54	1,8	85	25,9	7	TF	C-leg	1C40
TF4	M	43	1,6	72	26,7	3	KD	Rheo knee	Variflex
TF5	F	49	1,7	53	19,4	25	KD	Total Knee	Elation
TF6	M	44	1,7	47	16,6	18	TF	C-leg	Silhouette
TF7	F	26	1,7	65	23,9	2,5	Gritti	Rheo knee	Elation
TF8	M	26	1,8	56	17,3	1,5	TF	C-leg	Pro-Flex
TF9	M	32	1,8	95	29,3	7	TF	Rheo knee XC	Pro-Flex
TF10	M	35	1,7	83	29,1	9,5	KD	C-leg	Triton
Mean		41,5	1,73	68,8	23,0	10,5			
SD		11,3	0,07	15,2	4,7	10,6			

BMI, body mass index; F, female; M, male; TF, Transfemoral amputation; KD, Knee disarticulation, SD, standard deviation.

The prosthetic devices are from Ottobock (C-Leg, Triton, and 1C40) from Ossür (Rheo Knee, Mauch SNS, Total knee TK200, Variflex, Elation and Pro-Flex) and from Freedom Innovation (Silhouette).

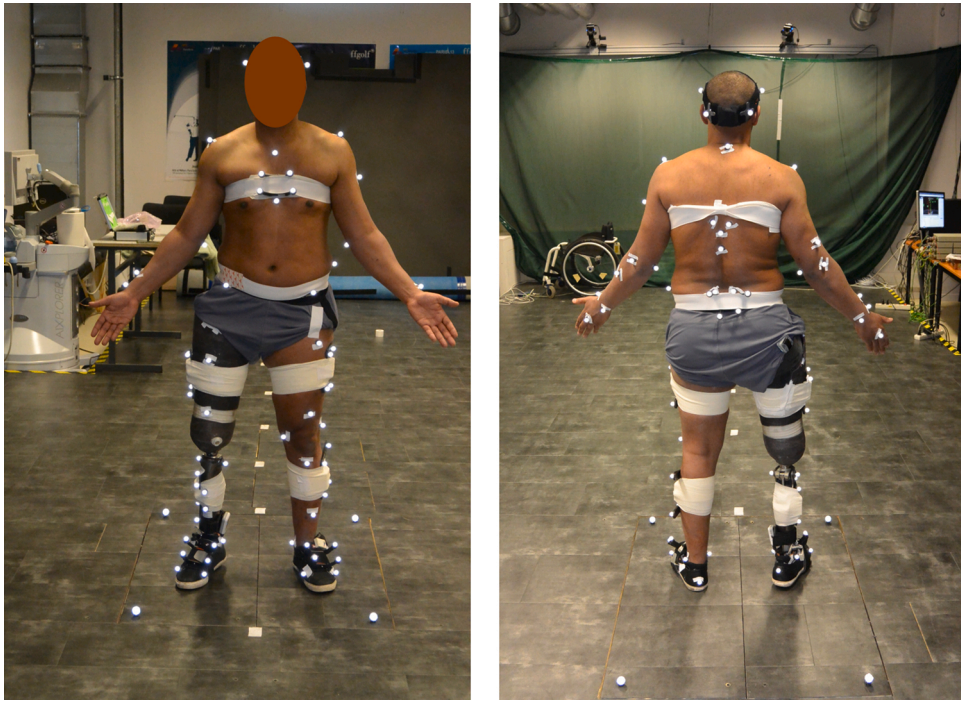


Fig. 1. A participant during the static standing posture. The 59-marker set can be appreciated on the front and back photographs.

2.3.2. Body segment networks

Following the contribution analysis, different body segment networks (BSNs) were proposed for the estimation of BCoM acceleration. The BSN-based BCoM acceleration was computed as the sum of the segmental contributions for the N included segments (Eq. (3)):

$$a_{BCoM, BSN} = \sum_{i=1}^N \text{Contrib}_{seg_i} = \sum_{i=1}^N \frac{m_{seg_i}}{\sum_{j=1}^N m_{seg_j}} a_{SCoM_i} \quad (3)$$

The inertial-model and BSN-based BCoM accelerations were compared to the reference BCoM acceleration over the central prosthetic gait cycle of each trial using Pearson's correlation coefficients (with an alpha-level of 0.05) as well as peak-to-peak normalized root-mean square errors (NRMSE, defined following [24]), subsequently averaged over all patients. The results achieved when including a single segment, namely the pelvis or the trunk, are also provided as an indication of the performance of the single-segment paradigm. Given the low number of available strides, only descriptive statistics were provided.

3. Results

A total of 25 complete prosthetic gait cycles were retrieved for the analysis, with an average of 3 gait cycles per participant (range 1–6).

3.1. Segmental contributions

Absolute and relative contribution weights are represented as stacked bar plots (Fig. 2) in order to observe both the weight of individual segments and between-segments compensations. For instance, contributions from the right and left upper limbs are shown to compensate each other in the anteroposterior direction (Fig. 2a.) and to account for less than 20 % of the 3D BCoM acceleration (Fig. 2d.–f.).

Similarities between SCoM and BCoM accelerations along the prosthetic gait cycle are reported in Fig. 3.

Overall, the trunk, pelvis and both thighs appear to be the main contributors of the BCoM acceleration in all three directions, both in terms of similarity and weight. Shank and foot segments being also significant contributors of the BCoM acceleration during the

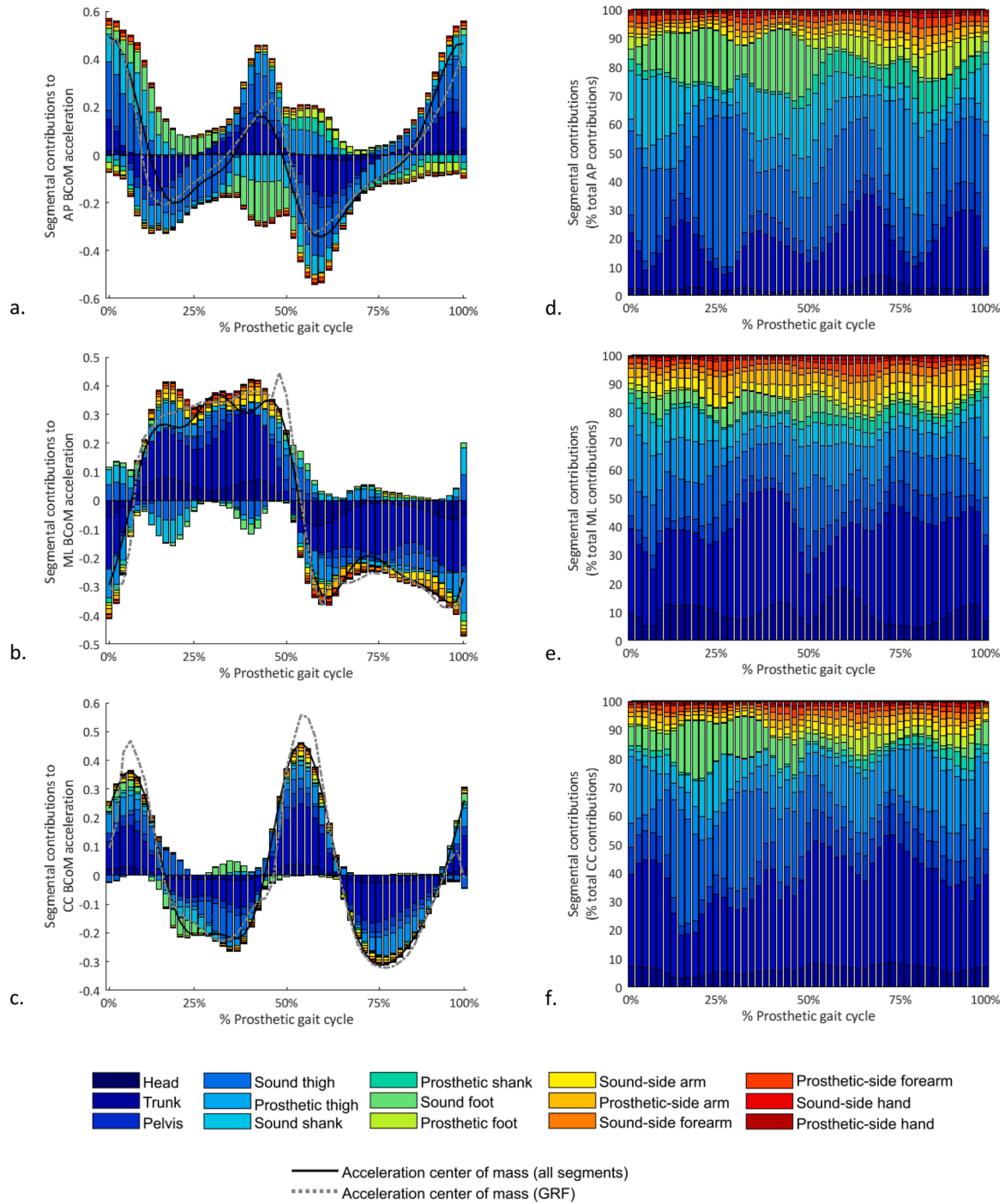


Fig. 2. Segmental contributions to the whole body center of mass (BCoM) acceleration compared to inertial-model based acceleration and reference acceleration (GRF) and total contributions in the anteroposterior [AP] direction (a. and d.), mediolateral [ML] direction (b. and e.) and in the vertical [V] direction (c. and f.) every 2% of the prosthetic gait cycle.

(a.–c.) Segmental contribution weights normalized per axial BCoM peak-to-peak acceleration; (d.–f.) Segmental contributions weights expressed as percent of total absolute contribution. (For a better interpretation of the figure, the reader is referred to the coloured web version of this article).

contralateral stance phase (Fig. 2d.,f.), the trunk, pelvis, and segments from both lower limb segments seem promising sensor locations for BCoM acceleration estimation.

3.2. Body segment networks

Several networks combining from three to six segment locations were considered for further analysis and compared to the reference BCoM acceleration along with the single-segment paradigm (Table 2). It

should be noted that the inertial-model-based BCoM acceleration achieved mean errors of $9.9 \pm 1.8 \%$, $10.7 \pm 2.4 \%$ and $10.6 \pm 2.4 \%$ in the anteroposterior, mediolateral and vertical directions respectively. The five-segment model including the trunk, thighs and feet exhibited the higher agreement with reference BCoM acceleration while the three-segment models including the trunk and either shanks or thighs represented the best compromises for 2D BCoM acceleration estimation in the sagittal and frontal plane, respectively. The estimated BCoM acceleration with the pelvis paradigm, and the BSN models including either the

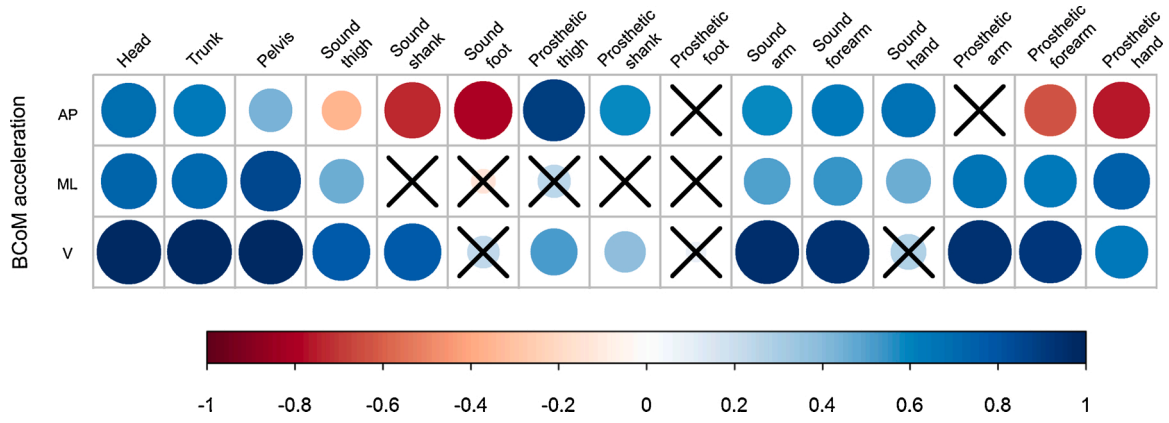


Fig. 3. Average correlation coefficients between each body segment centers of mass (SCoM) accelerations and inertial model-based body center of mass acceleration (BCoM) along the prosthetic gait cycle in the anteroposterior (AP), mediolateral (ML), and vertical (V) directions. Crossed correlations indicate that the correlation was non-significant (p -value > 0.05). The darker and bigger the circle, the stronger the correlation (blue tones: positive correlation, red tone: negative correlations). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

Table 2

Comparison of body center of mass (BCoM) acceleration derived from various body segment network (BSN) models to the reference acceleration from force plates.

Number of segments	Included segments	NRMSE (%)			Pearson's		
		AP	ML	V	AP	ML	V
1	Pelvis	25.3 (2.5)	26.1 (8.1)	11.1 (1.9)	0.65 (0.07)	0.60 (0.28)	0.91 (0.05)
1	Trunk	20.0 (2.8)	20.9 (3.2)	10.5 (2.0)	0.74 (0.08)	0.83 (0.07)	0.92 (0.04)
3	Pelvis, thighs	23.3 (3.4)	24.5 (6.6)	14.4 (2.8)	0.83 (0.04)	0.63 (0.18)	0.84 (0.09)
3	Trunk, thighs	18.1 (2.0)	12.8 (2.8)	11.3 (2.4)	0.86 (0.03)	0.91 (0.05)	0.90 (0.06)
3	Trunk, shanks	13.4 (2.8)	16.4 (3.5)	10.7 (2.3)	0.86 (0.06)	0.86 (0.07)	0.91 (0.04)
3	Trunk, feet	25.1 (4.1)	18.6 (3.7)	11.8 (2.1)	0.56 (0.12)	0.84 (0.09)	0.90 (0.04)
4	Trunk, pelvis, thighs	18.1 (2.0)	12.8 (2.7)	11.0 (2.4)	0.85 (0.03)	0.92 (0.05)	0.90 (0.06)
5	Trunk, thighs, shanks	13.9 (1.8)	13.4 (3.9)	11.3 (2.4)	0.92 (0.02)	0.91 (0.07)	0.90 (0.06)
5	Trunk, thighs, feet	11.6 (1.8)	12.7 (2.7)	10.7 (2.5)	0.92 (0.02)	0.92 (0.05)	0.91 (0.05)
6	Trunk, pelvis, thighs, shanks	13.5 (1.6)	13.6 (3.9)	11.0 (2.4)	0.92 (0.01)	0.91 (0.07)	0.91 (0.05)
6	Trunk, pelvis, thighs, feet	12.2 (2.1)	12.9 (2.8)	10.5 (2.5)	0.90 (0.03)	0.92 (0.05)	0.91 (0.05)

Results are provided as mean (standard deviation).

NRMSE = Normalized root mean square error; AP = Anteroposterior; ML = Mediolateral; V = Vertical.

trunk and shanks or the trunk, thighs and feet are represented in Fig. 4 against the reference BCoM acceleration.

4. Discussion

4.1. Segmental contributions

The first objective of the study was to quantify the segmental contributions to BCoM acceleration in people with transfemoral

amputation.

Similarly as in able-bodied gait [25], the trunk was found to be the major contributor to BCoM acceleration in the vertical and mediolateral directions while the thighs were the prime acceleration generator in the direction of progression. They were indeed found to explain more than 53 % of the BCoM acceleration, which might be due to their being the heaviest three segments of the body.

The analysis of trunk and pelvis contributions to BCoM acceleration is of particular interest as they are often used in the literature as proxy measures of BCoM motion [2,26]. Stronger correlations were found in the vertical direction ($r = 0.92$) than in the anteroposterior and mediolateral directions ($r \geq 0.74$ and $r \leq 0.65$ for the trunk and pelvis, respectively). This corroborates previous findings regarding the single-segment paradigm: it might be accurate enough for the study of vertical BCoM motion [26] while unsuited for 3D BCoM motion tracking [14,15].

Interestingly, contrary to what was observed in able-bodied subjects [10], segmental contributions were not observed to be near-constant during the prosthetic stance phase in the vertical direction (Fig. 2f.). An increased weight of the contribution of the sound leg accelerations is indeed observed at the beginning of the prosthetic cycle. This might result from sound-side ankle plantarflexion at push-off, which was shown to be a major determinant of vertical BCoM motion [27]. In the anteroposterior direction, the lower limbs were found to constitute the primary contributor to BCoM acceleration, with the sound limb accounting for almost half of the total BCoM acceleration. The asymmetry in the contribution weight might be partly explained by the lower mass of the prosthetic limb compared to the contralateral limb. Additionally, gait compensations involving the hip and ankle joints of the sound limb are frequent in this population [4,28] and may have led to increased accelerations of the contralateral thigh, shank and foot segments.

In light of these results, the inclusion of either shank or foot sensors in addition to that of the thighs seems relevant for BCoM acceleration estimation. This is also supported by the fact that shank sagittal angles were previously shown to predict BCoM displacement along with thigh and upper body segments in the asymptomatic population [29,30].

Similarly to [10], the head and upper limbs were discarded from the list of potential sensor locations for the wearable estimation of BCoM acceleration during straight level walking. Indeed, while an important contributor to BCoM mediolateral acceleration, head motion can often be uncorrelated from the whole-body motion [10]. Regarding the upper limbs, their minor contribution weight (< 10 % for each limb) was found to be near-constant along the gait cycle, possibly due to their reduced mass relative to the body's or to their limited range of motion during straight walking. However, the upper limbs may play a more important

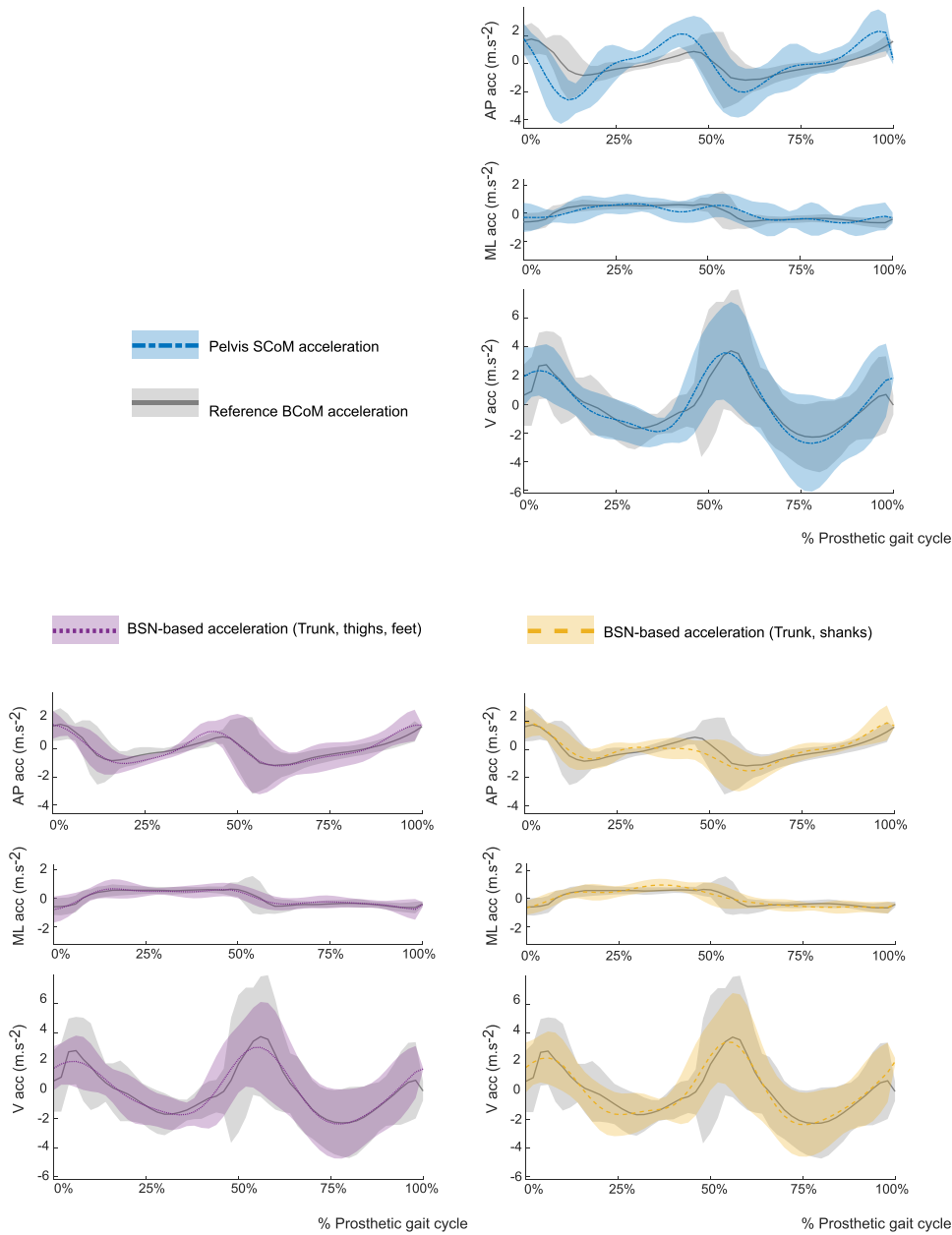


Fig. 4. Average acceleration (acc) of the body center of mass (BCoM) as estimated with the single-sensor paradigm (pelvis, blue dotted line), the body segment network (BSN) including the trunk and shanks segments (yellow dashed line), the BSN model including the trunk, thighs and feet segments (purple dotted line) compared to the reference force plates-based BCoM acceleration (gray straight line) in the anteroposterior (AP), mediolateral (ML) and vertical (CC) directions. Shaded regions represent the interval [mean - standard deviation, mean + standard deviation] for each estimate of the BCoM acceleration. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

role in other ambulation situations, especially when mitigating disturbances to avoid falling [31], and should therefore not be systematically discarded.

4.2. Body segment networks

Following the identification of the major contributors to BCoM acceleration, different BSNs were devised including from three to six segments with the aim of finding a compromise between accuracy and number of segments to be considered. The higher weight and agreement of trunk acceleration with the BCoM acceleration compared to the pelvis one (Figs. 2 and 3) favored the investigation of three and five-sensor models involving the trunk and lower-limb segments. Relevance of this choice was confirmed by the achieved results: poorer accuracy was achieved when using the pelvis in a single-segment paradigm compared to when using the trunk and adding the pelvis to a trunk-based BSN model improved the NRMSE by less than 0.5 % (Table 2). Thus, our results advocate for the inclusion of the trunk segment when tracking

BCoM motion in people with transfemoral amputation during straight walking. This is in agreement to previous literature reporting significant trunk 3D motion and pelvis rotations in this population [17].

The added-value of including several segments compared to the trunk-only or pelvis-only models for the anteroposterior and mediolateral components of BCoM acceleration is demonstrated in Table 2 and Fig. 4. Our results strengthen previous findings showing the unsuitability of the single-sensor paradigm for accurate capture of the 3D BCoM motion in asymmetrical gait [2,14–16,26]. In particular, the limited agreement and higher excursions of the BCoM acceleration estimated with a single segment compared to the reference BCoM acceleration in the anteroposterior direction might be due to an increased range of motion of the upper body in people with transfemoral amputation [17].

Following a similar procedure for the selection of segments in able bodies, Shahabpoor and coworkers developed an BSN model including the trunk, the pelvis and a thigh while accounting for the masses of all the body segments [10]. Their method required the computation of

SCoM accelerations cross-correlation matrices for each participant. In the present study, one 3-segment and both 5-segment BSN models achieved similar or improved accuracy compared to that reported by [10] with subject-specific matrices, without requiring the computation of subject-specific cross-correlation matrices. This is an important outcome, as it suggests that the 17-MIMU calibration procedure to obtain subject-specific cross-correlation matrix may not be necessary. The number of sensors included can be further reduced to three MIMUs on the trunk and shanks when aiming at capturing only the BCoM acceleration in the sagittal plane (Table 2). In order to further improve the compromise between accuracy and number of required sensors, an interesting track of research could be to propose kinematic models for groups of segments.

4.3. Limitations and sources of errors

The contribution analysis presented in this study was performed by comparing individual SCoM accelerations to that of the BCoM using the inertial model developed in [22]. Since inertial models were shown to influence the retrieved BCoM motion [2,32], alternative models could have led to different results. Nevertheless, the present analysis provided the same major contributors of BCoM acceleration as in the literature on able-bodied individuals [10,25], with specificities that are presumably related to the specific gait pattern of people with transfemoral amputation. It should be noted that the inertial model used achieved mean NRMSE of about 10 % in all directions with respect to force plate-based reference acceleration, which may explain why no further improvement in accuracy was achieved by the different BSN model-based estimations when adding segments, mainly in the vertical direction.

BSN models presented within this study were developed and validated with data derived from optical motion capture rather than wearable sensors. Therefore, their validity should be verified when using MIMUs. The latter provide raw acceleration and angular velocity measured at the origin and along the axes of the MIMU local frame, as well as orientation data in a global reference frame. To transfer the measured accelerations at the SCoM, the position of each MIMU relative to the underlying SCoM must be obtained and angular velocity differentiated. These processes may introduce errors which could jeopardize the accuracy of BCoM estimates [18] while the direct measurement of acceleration may reduce the errors due to the double differentiation of position data when using optoelectronic systems. Therefore, the investigation of the impact of MIMUs positioning relative to the underlying SCoM and the development of wearable methods allowing the identification of these relative positions represent research tracks of interest for the future. Furthermore, the orientation outputs estimated from MIMU signals have been shown to be affected by ferromagnetic perturbations [33], which may result from the ground within buildings or prosthetic components. In such conditions, different MIMUs may sense different Earth-fixed frames [33], which could introduce new errors when computing BCoM acceleration from a weighted average of estimated SCoM acceleration. However, similar accuracy levels achieved with a full body instrumentation using either MIMUs or optoelectronic systems in [18] is promising and suggests that transferring the BSN models developed within this study in a wearable framework might be feasible.

Finally, the suitability of the proposed BSN models to accurately capture the BCoM 3D velocity and displacement, which are relevant parameters for motion analysis in people with transfemoral amputation, should also be investigated in the future.

5. Conclusions

This study investigated the feasibility of estimating BCoM acceleration in people with transfemoral amputation from the acceleration of a limited number of segments during straight level walking. Including a minimum of five segments provides an accurate estimation of 3D BCoM acceleration compared to the literature, while only three segments are

necessary for the estimation of 2D acceleration. The trunk segment was shown to be crucial for the estimation of 3D BCoM acceleration and should be instrumented along with a minimum of two lower-limb segments.

The promising results achieved in this study may benefit the rehabilitation field by allowing the development of a wearable-sensors based framework for the obtention of BCoM-derived parameters for in-the-field patients monitoring.

CRediT authorship contribution statement

Emeline Simonetti: Conceptualization, Investigation, Methodology, Software, Formal analysis, Writing - original draft, Visualization. **Elena Bergamini:** Conceptualization, Writing - review & editing, Supervision. **Joseph Bascou:** Investigation, Writing - review & editing. **Giuseppe Vannozzi:** Writing - review & editing. **Hélène Pillet:** Conceptualization, Investigation, Writing - review & editing, Supervision, Project administration.

Declaration of Competing Interest

The authors report no declarations of interest.

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References

- [1] A.E. Minetti, C. Cisotti, O.S. Mian, The mathematical description of the body centre of mass 3D path in human and animal locomotion, *J. Biomech.* 44 (2011) 1471–1477, <https://doi.org/10.1016/j.jbiomech.2011.03.014>.
- [2] G. Pavel, E. Seminati, D. Cazzola, A.E. Minetti, On the estimation accuracy of the 3D body center of mass trajectory during human locomotion: inverse vs. forward dynamics, *Front. Physiol.* 8 (2017) 1–13, <https://doi.org/10.3389/fphys.2017.00129>.
- [3] L. Tesio, V. Rota, The motion of body center of mass during walking: a review oriented to clinical applications, *Front. Neurol.* 10 (2019) 1–22, <https://doi.org/10.3389/fneur.2019.00999>.
- [4] X. Bonnet, C. Villa, P. Fodé, F. Lavaste, H. Pillet, Mechanical work performed by individual limbs of transfemoral amputees during step-to-step transitions: effect of walking velocity, *Proc. Inst. Mech. Eng. Part H J. Eng. Med.* 228 (2014) 60–66, <https://doi.org/10.1177/0954411913514036>.
- [5] G.N. Askew, L.A. McFarlane, A.E. Minetti, J.G. Buckley, Energy cost of ambulation in trans-tibial amputees using a dynamic-response foot with hydraulic versus rigid “ankle”: insights from body centre of mass dynamics, *J. Neuroeng. Rehabil.* 16 (2019) 1–12, <https://doi.org/10.1186/s12984-019-0508-x>.
- [6] G. Pavel, F. Salis, A. Cereatti, E. Bergamini, Body center of mass trajectory and mechanical energy using inertial sensors: a feasible stride? *Gait Posture* 80 (2020) 199–205, <https://doi.org/10.1016/j.gaitpost.2020.04.012>.
- [7] A.L. Hof, M.G.J. Gazendam, W.E. Sinke, The condition for dynamic stability, *J. Biomech.* 38 (2005) 1–8, <https://doi.org/10.1016/j.jbiomech.2004.03.025>.
- [8] N. Al Abiad, H. Pillet, B. Watier, A mechanical descriptor of instability in human locomotion: experimental findings in control subjects and people with transfemoral amputation, *Appl. Sci.* 10 (2020), <https://doi.org/10.3390/app10030840>.
- [9] H. Pillet, X. Drevelle, X. Bonnet, C. Villa, N. Martinet, C. Sauret, J. Bascou, I. Loiret, F. Djian, N. Rapin, J. Mille, P. Thoreux, P. Fodé, J. Paysant, P. Guérit, F. Lavaste, APSIC: training and fitting amputees during situations of daily living, *IRBM* 35 (2014) 60–65, <https://doi.org/10.1016/j.irbm.2014.02.005>.
- [10] E. Shahabpoor, A. Pavic, J.M.W. Brownjohn, S.A. Billings, L.Z. Guo, M. Bocian, Real-life measurement of tri-axial walking ground reaction forces using optimal network of wearable inertial measurement units, *IEEE Trans. Neural Syst. Rehabil. Eng.* 26 (2018) 1243–1253, <https://doi.org/10.1109/TNSRE.2018.2830976>.

- [11] A. Ancillao, S. Tedesco, J. Barton, B. O'flynn, Indirect measurement of ground reaction forces and moments by means of wearable inertial sensors: a systematic review, *Sensors (Switzerland)* 18 (2018), <https://doi.org/10.3390/s18082564>.
- [12] E. Bergamini, G. Ligorio, A. Summa, G. Vannozzi, A. Cappozzo, A.M. Sabatini, Estimating orientation using magnetic and inertial sensors and different sensor fusion approaches: accuracy assessment in manual and locomotion tasks, *Sensors* 14 (2014) 18625–18649, <https://doi.org/10.3390/s141018625>.
- [13] L. Pacher, C. Chatellier, R. Vauzelle, L. Fradet, Sensor-to-segment calibration methodologies for lower - body kinematic analysis with inertial sensors: a systematic review, *Sensors* 20 (2020), <https://doi.org/10.3390/s20113322>.
- [14] A. Meichtry, J. Romkes, C. Gobelet, R. Brunner, R. Müller, Criterion validity of 3D trunk accelerations to assess external work and power in able-bodied gait, *Gait Posture* 25 (2007) 25–32, <https://doi.org/10.1016/j.gaitpost.2005.12.016>.
- [15] B. Jeong, C.Y. Ko, Y. Chang, J. Ryu, G. Kim, Comparison of segmental analysis and sacral marker methods for determining the center of mass during level and slope walking, *Gait Posture* 62 (2018) 333–341, <https://doi.org/10.1016/j.gaitpost.2018.03.048>.
- [16] M.I. Mohamed Refai, B.J.F. Van Beijnum, J.H. Buurke, P.H. Veltink, Portable gait lab: estimating 3D GRF using a pelvis IMU in a foot IMU defined frame, *IEEE Trans. Neural Syst. Rehabil. Eng.* 28 (2020) 1308–1316, <https://doi.org/10.1109/TNSRE.2020.2984809>.
- [17] H. Goujon-Pillet, E. Sapin, P. Fodé, F. Lavaste, Three-dimensional motions of trunk and pelvis during transfemoral amputee gait, *Arch. Phys. Med. Rehabil.* 89 (2008) 87–94, <https://doi.org/10.1016/j.apmr.2007.08.136>.
- [18] A. Karatsidis, G. Bellusci, H.M. Schepers, M. de Zee, M.S. Andersen, P.H. Veltink, Estimation of ground reaction forces and moments during gait using only inertial motion capture, *Sensors (Switzerland)* 17 (2017) 1–22, <https://doi.org/10.3390/s17010075>.
- [19] B. Fasel, J. Spörri, P. Schütz, S. Lorenzetti, K. Aminian, An inertial sensor-based method for estimating the athlete's relative joint center positions and center of mass kinematics in alpine ski racing, *Front. Physiol.* 8 (2017) 850, <https://doi.org/10.3389/fphys.2017.00850>.
- [20] A.G. Cutti, G. Verni, G.L. Migliore, A. Amoresano, M. Raggi, Reference values for gait temporal and loading symmetry of lower-limb amputees can help in refocusing rehabilitation targets, *J. Neuroeng. Rehabil.* 15 (2018), <https://doi.org/10.1186/s12984-018-0403-x>.
- [21] V. Agrawal, R. Gailey, C. O'Toole, I. Gaunaud, T. Dowell, Symmetry in external work (SEW): a novel method of quantifying gait differences between prosthetic feet, *Prosthet. Orthot. Int.* 33 (2009) 148–156, <https://doi.org/10.1080/03093640902777254>.
- [22] H. Pillet, X. Bonnet, F. Lavaste, W. Skalli, Evaluation of force plate-less estimation of the trajectory of the centre of pressure during gait. Comparison of two anthropometric models, *Gait Posture* 31 (2010) 147–152, <https://doi.org/10.1016/j.gaitpost.2009.09.014>.
- [23] R. Dumas, L. Chèze, J.P. Verriest, Adjustments to McConville et al. and Young et al. body segment inertial parameters, *J. Biomech.* 40 (2007) 543–553, <https://doi.org/10.1016/j.jbiomech.2006.02.013>.
- [24] L. Ren, R.K. Jones, D. Howard, Whole body inverse dynamics over a complete gait cycle based only on measured kinematics, *J. Biomech.* 41 (2008) 2750–2759, <https://doi.org/10.1016/j.jbiomech.2008.06.001>.
- [25] C. Gillet, J. Duboy, F. Barbier, S. Armand, R. Jeddi, F.X. Lepoutre, P. Allard, Contribution of accelerated body masses to able-bodied gait, *Am. J. Phys. Med. Rehabil.* 82 (2003) 101–109, <https://doi.org/10.1097/00002060-200302000-00004>.
- [26] S.A. Gard, S.C. Miff, A.D. Kuo, Comparison of kinematic and kinetic methods for computing the vertical motion of the body center of mass during walking, *Hum. Mov. Sci.* 22 (2004) 597–610, <https://doi.org/10.1016/j.humov.2003.11.002>.
- [27] C. Hayot, S. Sakka, P. Lacouture, Contribution of the six major gait determinants on the vertical center of mass trajectory and the vertical ground reaction force, *Hum. Mov. Sci.* 32 (2013) 279–289, <https://doi.org/10.1016/j.humov.2012.10.003>.
- [28] X. Drevelle, C. Villa, X. Bonnet, I. Loiret, P. Fodé, H. Pillet, Vaulting quantification during level walking of transfemoral amputees, *Clin. Biomech. (Bristol, Avon)* 29 (2014) 679–683, <https://doi.org/10.1016/j.clinbiomech.2014.04.006>.
- [29] D. Arumukhom Revi, A.M. Alvarez, C.J. Walsh, S.M.M. De Rossi, L.N. Awad, Indirect measurement of anterior-posterior ground reaction forces using a minimal set of wearable inertial sensors: from healthy to hemiparetic walking, *J. Neuroeng. Rehabil.* 17 (2020) 82, <https://doi.org/10.1186/s12984-020-00700-7>.
- [30] D.S. Mohan Varma, S. Sujatha, Segmental contributions to the center of mass movement in normal gait, *Appl. Math. Model.* 46 (2017) 328–338, <https://doi.org/10.1016/j.apm.2017.01.075>.
- [31] M. Popović, A. Hofmann, H. Herr, Angular Momentum Regulation During Human Walking: Biomechanics and Control, vol. 3, 2004, pp. 2405–2411, <https://doi.org/10.1109/robot.2004.1307421>.
- [32] R.D. Catena, S.H. Chen, L.S. Chou, Does the anthropometric model influence whole-body center of mass calculations in gait? *J. Biomech.* 59 (2017) 23–28, <https://doi.org/10.1016/j.jbiomech.2017.05.007>.
- [33] P. Picerno, A. Cereatti, A. Cappozzo, A spot check for assessing static orientation consistency of inertial and magnetic sensing units, *Gait Posture* 33 (2011) 373–378, <https://doi.org/10.1016/j.gaitpost.2010.12.006>.