# Development of a Wearable Framework for the Assessment of a Mechanical-Based Indicator of Falling Risk in the Field

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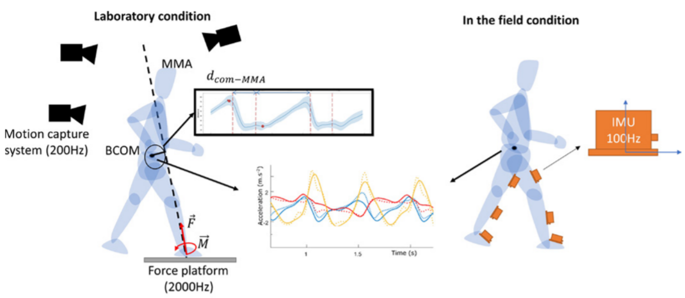
## HIGHLIGHTS

- Gait instability can be quantified by the distance between the BCoM and the Minimal Moment Axes.
- The kinematics of the BCoM can be determined with 6 wearable IMUs fixed at the lower limbs.
- It should be possible to quantify the gait instability in the field with IMUs.

*Keywords:*
Locomotion
Gait instability
Wearable sensors
Falling risk

## GRAPHICAL ABSTRACT




Our research compares kinematics of the Body Center Of Mass (BCOM) during gait obtained in the lab with motion capture system and in the field with IMU. Moreover we define a new instabiblity parameter (distance from BCOM and the Mimimum Moment Axis - $d_{com-MMA}$). We obtain the range of this parameter for normal healthy condition

## ABSTRACT

*Objectives:* The characterization of the instability of gait is a current challenge of biomechanics. Indeed, risks of falling naturally result from the difficulty to control perturbations of the locomotion pattern. Hence, the assessment of a synthetic parameter able to quantify the instability in real time will be useful for the prevention of falls occurring in this context. Thus, the objective of the present study, in two steps, was to propose and evaluate a relevant parameter to quantify the risk of fallings.

*Material and Methods:* Experimental analysis of the gait of 11 able-bodied subjects from a motion capture system in laboratory condition was performed. The distance of the Body Center of Mass (BCOM) to the Minimal Moment Axis (MMA) was computed as a proxy of whole-body angular momentum variations. In a second step, we quantified the kinematics during gait with wearable Inertial Measurement Units (IMU) fixed on two individuals (one able bodied person and one person with transfemoral amputation). We compared the IMU-based BCOM kinematics with a motion capture reference system to verify the accuracy of our measures in the field.

*Results:* Normative thresholds of the distance of the Body Center of Mass (BCOM) to the Minimal Moment Axis (MMA) during able-bodied level walking were assessed. The average error between the BCoM displacement computed from the IMU and from the reference vicon data of 4 mm, 3 mm and 53 mm on the mediolateral, anteroposterior and vertical axes respectively.

*Conclusion:* All these results make it possible to consider the determination of the risks of falls in the field at mid-term. the research on an optimal configuration that maintain the performance while simplifying the device will be essential to make it acceptable by the individuals.



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## 1. Introduction

Falling risks can result from different origins but a lot of falls occur during locomotion. The instability of gait that can be defined as the inability to control gait deviations due to perturbations is therefore a determinant of the immediate risk of falling. Thus, quantifying the instability of gait is of crucial importance for the monitoring of falling risk. In the literature, variation of the whole-body angular momentum has been identified as a good mechanical candidate to this aim [1]. Indeed, when modeling the human body as a polyarticulated chain of rigid segments [2], the second vectorial equation from dynamic laws links at a given point the variation of the angular momentum to the sum of the moments of external forces. Computing this relation at the Body Center of Mass (BCoM) makes sense to understand how external mechanical actions modify the rotation around the BCoM. Thus, an increase of the variation of angular momentum at BCoM can be interpreted as a perturbation that must be limited by the person to avoid uncontrolled rotational motion around the center of gravity. If the variation of the angular momentum is therefore a meaningful parameter from a mechanical point of view, its interpretation can be a bit challenging in a clinical context and requires specific knowledge of mechanical concepts. Yet, to propose an alternative indicator that can be monitored and understood by clinicians or patients, the distance between the BCoM and the Minimal Moment Axis (MMA) referred hereafter as $d_{BCOM-MMA}$ has been provided in the literature [3,4]. Its evolution is directly reated to the whole body angular momentum with the advantage of being easy both to understand and represent.

In the same time, the way the indicator can be obtained is very important when dealing with real time monitoring. The 3D kinetics of the BCoM are usually acquired in laboratory conditions with either force plates or the combination of inertial models and optical motion capture system (OMCS), generally not available in clinical routine (cost, set-up time, ...). Therefore, the use of magneto-inertial measurement units (MIMUs) has been proposed as a wearable alternative for the estimation of BCoM-derived parameters. Shahabpoor et al. proposed a method to select a reduced number of MIMUs for the estimation of vertical BCoM acceleration [5]. This method has been recently extended and validated considering the 3D nature of BCoM movement to obtain accurate 3D BCoM acceleration of one person with transfemoral amputation [6].

The aim of the present article is (i) to give the normative thresholds of $d_{BCOM-MMA}$ associated to able-bodied level walking (ii) to describe a framework that could be used to assess $d_{BCOM-MMA}$ from fully wearable sensors and to present bottlenecks that still have to be overcome for the final implementation of this framework.

## 2. Material and method

### 2.1. Theoretical background

Contact mechanical actions acting on a rigid body can be represented at any point $A$, by a force $\boldsymbol{F}$ and a moment $\boldsymbol{M_A}$. $\boldsymbol{F}$ and $\boldsymbol{M_A}$ define a moment field that can be expressed at any point B as:

$$\boldsymbol{M_B} = \boldsymbol{M_A} + \boldsymbol{F} \times \boldsymbol{P}_{(\boldsymbol{A},\boldsymbol{B})} \tag{1}$$

Where $\times$ denotes the cross product and $\boldsymbol{P}_{(\boldsymbol{A},\boldsymbol{B})}$ is the position of $B$ with respect to $A$.

There exists one axis called the Minimal Moment Axis (MMA) such that, at each point of this axis, the norm of the moment is minimal. Moreover, it can be demonstrated that this axis is collinear to $\boldsymbol{F}$ [7].

Assume that point $Q \in \mathrm{MMA}$.

MMA is computed as in [3]:

$$\forall B \text{ in space},\ \forall Q \in \mathrm{MMA}, \quad \boldsymbol{P}_{(B,Q)} = \frac{\boldsymbol{F} \times \boldsymbol{M_B}}{\|\boldsymbol{F}\|^2} + \lambda \boldsymbol{F}, \quad \lambda \in \mathbb{R}. \tag{2}$$

Considering the contact mechanical actions between the foot and the ground during gait and if we now assume that B is the BCoM

$$\forall Q \in \mathrm{MMA}, \boldsymbol{P}_{(\mathrm{BCoM},Q)} = \frac{\boldsymbol{F} \times \boldsymbol{M}_{\mathrm{BCoM}}}{\|\boldsymbol{F}\|^2} + \lambda \boldsymbol{F}, \lambda \in \mathbb{R}. \tag{3}$$

When $\lambda = 0$, Q is the orthogonal projection of the BCoM onto the *MMA*, and the distance $\left\|\boldsymbol{P}_{(\mathrm{BCoM},Q)}\right\|$ is the minimal distance between MMA and the BCoM. The vector $\boldsymbol{P}_{(\mathrm{BCoM},Q)}$ will be called $\boldsymbol{d}_{\mathrm{BCoM-MMA}}$ hereafter:

$$\boldsymbol{d}_{\mathrm{BCoM-MMA}} = \frac{\boldsymbol{F} \times \boldsymbol{M}_{\mathrm{BCoM}}}{\|\boldsymbol{F}\|^2} \tag{4}$$

From equation (4), we compute the Euclidian norm of the distance from BCoM position and force platform data:

$$d_{BCoM-\mathrm{MMA}} = \|\boldsymbol{d}_{\mathrm{BCoM-MMA}}\| = \frac{\|\boldsymbol{F} \times \boldsymbol{M}_{\mathrm{BCoM}}\|}{\|\boldsymbol{F}\|^2} \tag{5}$$

Interestingly, it can be noted that this distance can also be linked to the angular momemtum at the BCoM by considering the relation: $\dot{\boldsymbol{h}}_{\mathrm{BCoM}} = \boldsymbol{M}_{\mathrm{BCoM}}$ where $\dot{\boldsymbol{h}}_{\mathrm{BCoM}}$ is the derivative of the angular momentum. Thus:

$$\|\boldsymbol{d}_{BCoM-\mathrm{MMA}}\| = \frac{\left\|\dot{\boldsymbol{h}}_{\boldsymbol{COM}}\right\| \cdot |\sin\theta|}{\|\boldsymbol{F}\|} \tag{6}$$

With $\theta$ is the angle between $\boldsymbol{F}$ and $\dot{\boldsymbol{h}}_{\boldsymbol{COM}}$.

The components $\boldsymbol{d}_{\mathrm{BCoM-MMA}}$ in the global reference frame can also be computed and are denoted as $d_{ML}$, $d_{AP}$, $d_V$ on the Medio-Lateral, Anterior-Posterior and Vertical axes respectively. Each of these components can be related to the $\boldsymbol{M}_{\mathrm{BCoM}}$ components and therefore be interpreted in term of whole body angular momentum variation around the three axis.

$$\begin{bmatrix} M_{ML} \\ M_{AP} \\ M_V \end{bmatrix} = \begin{bmatrix} d_{ML} \\ d_{AP} \\ d_V \end{bmatrix} \times \begin{bmatrix} F_{ML} \\ F_{AP} \\ F_V \end{bmatrix} = \begin{bmatrix} F_V \times d_{AP} - F_{AP} \times d_V \\ F_{ML} \times d_V - F_V \times d_{ML} \\ F_{AP} \times d_{ML} - F_{ML} \times d_{AP} \end{bmatrix}$$

Assuming that during gait $F_V$ is the largest component, important values of $d_{AP}$ and $d_{ML}$ lead to high values of $M_{ML}$ and $M_{AP}$ and therefore important angular acceleration of the whole body respectively around the Medio-Lateral and the Antero-Posterior axes.

### 2.2. Experimental procedure

From the theoretical framework, it can be seen that the computation of $d_{COM-\mathrm{MMA}}$ necessitates to acquire the external mechanical action acting on the foot from the ground together with the current position of the BCoM. For ground reaction forces, the gold standard is the use of force plates that are integrated in the ground. For body center of mass kinematics, as stated in the introduction, motion capture can be used.

Thus, the first part of the present study, under approval of the ethics committee, included 11 able-bodied volunteers whose gait was analyzed using a motion capture system [3]. Each volunteer was equipped with 47 retro-reflective skin markers tracked at 200 Hz using a 13 Vicon motion capture system (Oxford Metrics). Ground reaction forces and moments (GRF) were simultaneously measured at 2000 Hz by three force plates (AMTI) and filtered with a 4th order Butterworth low-pass filter with a 15 Hz cut-off frequency. Each subject performed several trials at self-selected walking velocity.

In an attempt to monitor $d_{COM-\mathrm{MMA}}$ in real time, it is essential to compute previously mentioned input data from wearable sensors. Thus, on one hand, the kinematics of the center of mass has to be obtained from MIMUs distributed on the body. On the other hand, the 3D ground reaction forces and moments should be estimated at the center of pressure by considering simultaneously the evaluation of the total ground reaction forces from second Newton's law and the measurement of the estimation of the center of pressure from pressure insoles. In this methodology, only the moment around the vertical axis will not be retrieved but it has been shown to be low and should contribute for a small part to the moment generated at the BCoM.

Assessment of body center of mass kinematics from wearable sensor is still under development but in the present study the feasibility of such assessment is also presented. The developed framework is based on the methodology described by Simonetti et al. [6] to obtain the 3D acceleration of the BCoM. Briefly three steps are necessary: $i$) computing the 3D acceleration of each segment's center of mass (sCoM) from MIMUs data, $ii$) merging sCoM accelerations in a consistent common global frame $\boldsymbol{R_G}$, and $iii$) estimating the 3D BCoM acceleration from a weighted average of selected sCoM accelerations. Adapting the approach from Shahabpoor et al. [5], the trunk, pelvis, thighs and feet were identified as the major contributors in 3D BCoM acceleration in both able-bodied and people with amputation. Therefore, 6 MIMUs, manually aligned with the longitudinal axes of the segments, are adopted. For further details, the reader can refer to Simonetti et al. [6]. From the computation of 3D BCoM acceleration, 3D BCoM velocity and position can be retrieved from an integration process assuming integration constants. The feasibility of such approach has already been demonstrated [8].

One male individual with transfemoral amputation (mass: 83 kg, stature: 1.69 m) and one female able-bodied subject (mass: 56 kg, stature: 1.69 m) gave their written informed consent to participate in the study. They were instrumented with a full-body marker set [4] and 6 MIMUs (Xsens, 100 Hz) on the feet, thighs, pelvis, and trunk, each mounted on a 3D-printed plastic support with housings for 4 reflective markers. A motion capture system (VICON, 200 Hz) recorded the markers' positions while 4 photographs (front, back, both sides) were taken. Then, starting from a static standing posture, the participant walked at self-selected speed along an 8 m pathway, with 3 force plates (AMTI, 1000 Hz) in the middle. Synchronization between instruments was achieved by a trigger signal. Only one trial was acquired with three successive foot contacts on the force plates (i.e. a complete stride), and considered for analysis.

### *2.3. Data processing*

From motion capture data and the ground reaction force measured for force plates, the MMA along which the moment of the external contact forces is minimal was obtained. 3D BCoM position was determined from a regression method [9]. The vector $\boldsymbol{d}_{BCoM-MMA}$ between the BCoM and its projection on the MMA was computed at each instant of time (4), the norm of this vector $d_{BCoM-MMA}$ and its components in the global reference frame denoted as $d_{ML}$, $d_{AP}$, $d_V$ on the Medio-Lateral, Anterior-Posterior and Vertical axes respectively.

From wearable sensors, data were filtered using a zero-phase 4$^{\text{th}}$-order Butterworth filter (cut-off frequencies identified using a trial-and-error approach: 5 Hz markers and MIMUs, 10 Hz force plates). Reference BCoM acceleration was computed from force plates' signals. The agreement of both signals (IMU based vs reference) was quantified from Normalized Root Mean Square (in %) over the total acquisition time and $\rho$ coefficient of correlation for each component of the acceleration. Then, the IMU-based acceleration was integrated by direct integration to compute the cyclical part of the BCoM velocity. Average velocity (computed from [10]) was added to this cyclical part to obtain the instantaneous velocity of the BCoM. The signal was integrated a second time in the same condition to obtain the displacement of the BCoM from its initial position at the beginning of the trial. This displacement was compared to the displacement measured on the three axis of the reference frame from Vicon data.

**Table 1**
Average and standard deviation for the 11 participants of the ranges of $d_{ML}$, $d_{AP}$, $d_V$ and $d_{BCoM-MMA}$ in mm.

| Range (mm) | $d_{AP}$ | $d_{ML}$ | $d_V$ | $d_{BCoM-MMA}$ |
|---|---|---|---|---|
| Average | 42 | 101 | 7 | 70 |
| Standard deviation | 14 | 14 | 2 | 9 |

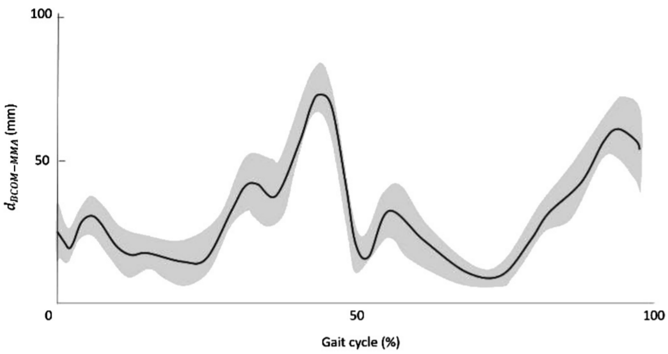


**Fig. 1.** Evolution of $d_{BCoM-MMA}$ over a gait cycle (average and corridor +/−1 standard deviation for the 11 participants).

## 3. Results

### *3.1. Normative thresholds of the distance of the minimal moment axis to the body center of mass during able bodied gait*

The results confirm that the components of the vector between the BCoM and its projection on the MMA remains low for able-bodied subjects walking on level ground (Table 1).

The pattern of $d_{BCoM-MMA}$ (Fig. 1) also shows that the norm increases at the end of the single support phase.

### *3.2. Preliminary results for the assessment of BCoM kinematics from wearable sensors*

High consistency between reference and MIMU 3D BCoM acceleration patterns can be observed (Fig. 2). The metrics confirmed the agreement between both signals.

The double integration of BCoM acceleration from MIMUs is represented on the Fig. 3 and compared to BCoM displacement obtained from VICON motion capture system. The numerical integration resulted in an expected drift that led to an average error between the BCoM displacement computed from the MIMU and from the reference vicon data of 4 mm, 3 mm and 53 mm on the mediolateral, anteroposterior and vertical axes respectively. The error was maximal at the end of the trial on the anteroposterior axis due to the accumulated drift and reached up to 106 mm whereas it remains below 19 mm on the mediolateral axis and 9 mm on the vertical one.

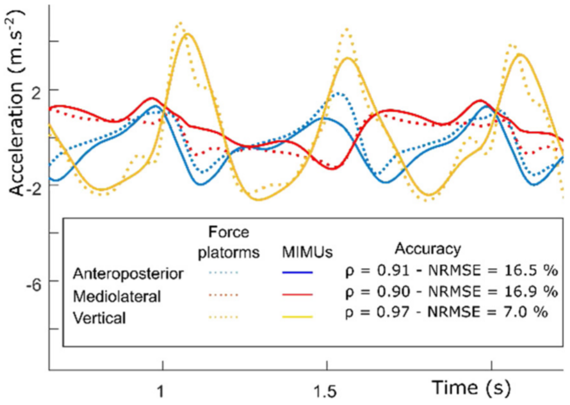


**Fig. 2.** BCoM acceleration obtained with force plates (dotted lines) and MIMUs (straight lines).

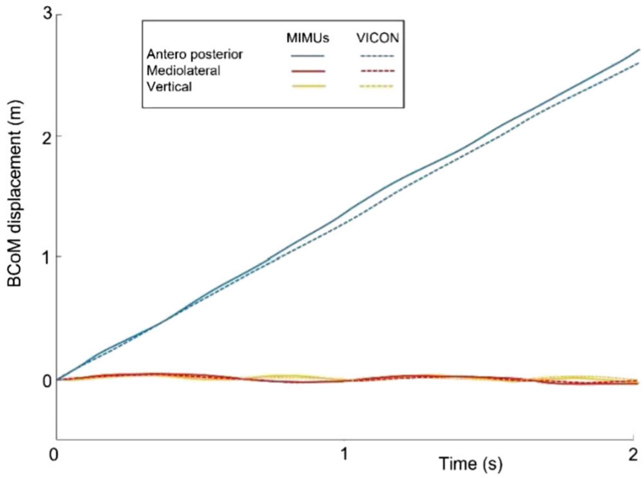


**Fig. 3.** BCoM displacement from its initial position obtained from reference VICON motion capture system (dotted lines) and MIMUs (straight lines).

## 4. Discussion

The aim of the present article was (i) to give the normative thresholds of $d_{BCOM-MMA}$ associated to able-bodied level walking (ii) to describe a framework that could be used to assess $d_{BCOM-MMA}$ from fully wearable sensors and to present bottlenecks that still have to be overcome for the final implementation of this framework.

The present article gives two important results. First, reference values of the distance of the Body Center of Mass (BCOM) to the Minimal Moment Axis (MMA) during able-bodied gait are computed and constitute the normative thresholds that could be used to characterize pathological or instable gait. The results are consistent with existing literature [3,4] and complete the values available from previous studies. On the Fig. 1, it can be noticed that the distance remains very low (under 5 cm) during the single support phase. Indeed during this phase, the person is in stance on one limb making it very vulnerable to unexpected perturbation. The low value is therefore consistent with the idea that the distance has to be controlled in situation at risk. On the contrary, the value increases up to 8 cm during double limb support. During this phase, it is logical that the variation of whole body angular momentum at the center of mass increases as the body has to tilt to pass from the trailing leg to the leading leg and to reorient the trajectory of the center of mass. In that sense, the increase of the distance must be higher but still be controlled. Thus, the values presented in the present study could serve as targets that could be implemented in monitoring tools useful for rehabilitation or as thresholds to detect immediate risk of falling. These reference values are also essential to fix the necessary accuracy in the determination of the distance, which is directly related to the one of the body center of mass trajectory.

As concerns the range of the components of $d_{BCOM-MMA}$ reported in Table 1, it can be noticed that $d_{AP}$ and $d_{ML}$ present the largest amplitude compared to $d_V$. Considering the link with the values of the moment at the center of mass recalled in the theoretical background part, these values could be used to dissociate the risk of falling backward or foreward and the risk of falling on each of the side.

Second, the potential of a fully wearable framework has been demonstrated based on a preliminary experiment with one individual during a single trial that revealed very encouraging results for the assessment of 3D BCoM kinematics (Fig. 2). This evaluation is a first step to prove the feasibility of such methodology but a lot of work remains in the development of a fully wearable procedure. Particularly, in the present study, the optoelectronic data were used at different steps detailed in [6]. Another limitation is that the hypothesis used in the study imposes to walk on level ground and will not be valid anymore in other walking conditions. Also, only the displacement of the BCoM from the initial position could be assessed in the absence of a mean to obtain its position directly from wearable sensors (Fig. 3). In the future, this could be overcome by combining the present model with a kinematic chains as already proposed in [11]. The kinematic chain should allow to initialize an approximated position of the BCoM essential to compute the distance to the MMA. To determine the distance of the Minimal Moment Axis to the body Center Of Mass, an estimation of the center of pressure would also be necessary and could be provided by wearable insoles. Finally, the obtention of BCoM trajectory and the synchronization with pressure insoles to estimate the position of the Minimal Moment Axis will be necessary. The accuracy of these parameters remains also to be investigated.

## 5. Conclusion

To conclude, the present article opens very important perspectives to compute an original mechanical indicator of the gait instability from wearable technologies. The results obtained are in favor of the feasibility of an ambulatory device to monitor this instability but some obstacles remain to be tackled. In addition to the computation of ground reaction forces and moments, the identification of decision thresholds is essential and should discriminate at risk situations avoiding false positive and false negative detections. This can only be based in extensive databases that should be acquired on frail and healthy control persons. Complementary experiments on biped robots and situations simulating instability could also be set. The advantage of a wearable device will be to adapt this threshold retroactively as long as data are acquired in populations at risk of falling. From the data acquired in the field, it will be possible not only to strengthen the performance of the indicator but also to increase the acceptability of the device by selecting the optimal sensor position, which could allow to minimize their numbers. Thus, the research on an optimal configuration that maintain the performance while simplifying the device will be essential to make it acceptable by the individuals

## Human and animal rights

The authors declare that the work described has been carried out in accordance with the Declaration of Helsinki of the World Medical Association revised in 2013 for experiments involving humans as well as in accordance with the EU Directive 2010/63/EU for animal experiments.

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## Author contributions

All authors attest that they meet the current International Committee of Medical Journal Editors (ICMJE) criteria for Authorship.

## CRediT authorship contribution statement

**H. Pillet:** Conception and design of study, acquisition of data, analysis and/or interpretation of data, drafting the manuscript, approval of the version of the manuscript to be published. **B. Watier:** Conception and design of study, acquisition of data, analysis and/or interpretation of data, revising the manuscript critically for important intellectual content, approval of the version of the manuscript to be published.

## Declaration of competing interest

The authors declare that they have no known competing financial or personal relationships that could be viewed as influencing the work reported in this paper.

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