# Can handling a weapon make soldiers more unstable?

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# Can handling a weapon make soldiers more unstable?

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**Abstract:** Gait stability in soldiers can be affected by task constraints that may lead to injuries. This study determined the effects of weapon handling and speed on gait stability in seventeen soldiers walking on a treadmill with and without a replica weapon at self-selected (SS), 3.5 km□h$^{-1}$, 5.5 km□h$^{-1}$, and 6.5 km□h$^{-1}$ while carrying a 23-kg load. Local dynamic stability was measured using accelerometry at the sacrum ($LDE_{SAC}$) and sternum ($LDE_{STR}$). No significant weapon and speed interaction were found. A significant effect of speed for the $LDE_{SAC}$, and a significant effect of speed and weapon for the $LDE_{STR}$ were found. Per plane analyses showed that the weapon effect was consistent across all directions for the $LDE_{STR}$ but not for $LDE_{SAC}$. Weapon handling increased trunk but did not affect pelvis stability. Speed decreased stability when walking slower than SS and increased when faster. These findings can inform injury prevention strategies in the military.

**Keywords:** load carriage; weapon handling; Lyapunov; stability

**Practitioner Summary:**

We determined the effects of two constraints in soldier's walking stability, weapon handling and speed, measured at the trunk and sacrum. No constraints interactions were found, however, lower stability when walking slow and greater stability with the weapon at the trunk can inform preventive strategies in military training.

## 1. Introduction

Carrying load is an essential component of military operations, which includes carrying weapons while in combat and during approach marches[1]. Soldiers' locomotor performance during load carriage can be affected by several factors including speed, load, type of armour, and load placement [2]. Load carriage in the military has been widely investigated from a biomechanical perspective with the aim of not only improving performance, but also to prevent the high incidence of musculoskeletal injuries[3]. For example, walking and training while carrying loads have been identified as modifiable and preventable non-battle causes of fall-related injuries[4-6]. Although spatiotemporal gait changes (e.g. stride length reductions) have been previously reported [7], a recent review concluded that there is minimal effect of load on these measures and that lower limb and trunk kinematic and kinetic adaptations are consistent across studies [8].

Fighting loads of 27-36 kg are recommended [1] for load optimization, of which the weapon is not a significant proportion [9]. Weapon carriage displaces the human-weapon-system centre-of-mass (CoM) forward[10] while also restricting the arms' natural swing, leading to increased in-phase pelvis-trunk coordination and its variability in the transverse plane, indicating an "*en bloc*" and variable rotatory movements [9]. Further, most gait kinetic changes have been associated with arm-swing restriction rather than the added mass during weapon handling[11]. Studies exploring the effects of arm swing have found that active swinging increases stability of the trunk when compared with normal and restricted arm swing, particularly in the mediolateral (ML) direction [12-14]. Although restricting arm swing does not decrease stability during walking, it may impair the ability to recover from an external perturbation[15], which may increase subsequent risk of falling or injury.

In addition to restricted arm swing, walking speed also exerts changes in walking patterns while carrying loads [7] and can be a major confounder when comparing groups or conditions. Speed, load, and their interaction have been found to significantly affect whole-body CoM and joint stiffness when walking [16]. Joint stiffness has been suggested to be a mechanism by which CoM excursion is controlled, which most likely improves the stability of walking. Gait stability can be assessed using several measures, however, using the local divergence exponent (LDE; $\lambda$) seems to be a more valid and sensitive method than, for example, step width variability [15,17]. The LDE "quantifies the average logarithmic rate of divergence of a system after a small perturbation"[15]. In other words, larger LDE values indicate more divergence and lower local dynamic stability, which has been associated to a decreased neuromuscular control of walking and increased risk of falling in older adults and neurological populations[18,19]. The LDE belongs to a set of measures aimed at quantifying the dynamic behavior of a system over time, which cannot be captured by linear metrics such as the coefficient of variation or standard deviation [20,21].

Studies of LDE have found that walking with relative (20% and 40% of body mass (BM)) and absolute (8.5 kg and 20 kg) loads lead to lower CoM stability in the ML direction[17]. Conversely, LDE of the CoM in the ML direction was found to be significantly lower when carrying a backpack which was associated with a less stable behaviour of motor outputs, when compared with an unloaded condition [22]. However, to date, no studies have explored the effects of weapon handling on local dynamic stability while walking at different speeds.

Assessing the stability of walking is paramount to understand the capacity of the neuromuscular system to deal with internal and external perturbations[21]. Weapon handling, speed, and load carriage impose external constraints that may lead to decreased gait stability and potentially increased risk of falling and injuries[23,24] during walking and training activities in the military [4]. In this regard, non-linear analysis of gait measures is a potentially useful tool not only in the assessment but also in the risk reduction of musculoskeletal injuries [25]. Non-linear analysis could assist instructors and clinicians to improve training programs to reduce the occurrence of instability-associated injuries. Therefore, the primary aim of this study was to determine the effects of weapon handling and walking speed on gait stability in soldiers. Since previous studies have found per-direction differential effects of external constraints on walking stability[14,17,22,26], a secondary aim was to determine whether weapon handling and walking speed effects are similar across different directions of motion.

## 2. Materials and Methods

### *2.1. Participants*

Seventeen active-duty Australian Army soldiers (5 females, 12 males; age: 25 ± 6 years; height: 177.6 ± 9.3 cm; mass: 80.7 ± 15.6 kg; military experience: 22.6 ± 21.3 months [13.0 – 32.1 95% CI range]) with no history of musculoskeletal or neurological injury in the six months prior to data collection participated in this study. Thirteen participants were trainees from the Australian Army School of Signals, whereas the other four participants were qualified soldiers from the Australian Army School of Artillery. The study was conducted according to the guidelines of the Declaration of Helsinki and approved by the Departments of Defence and Veterans' Affairs Human Research Ethics Committee and La Trobe University (302-20 and 02-2021, respectively).

*2.2. Walking tasks*

Participants performed eight 12-minute walking trials on an instrumented treadmill (AMTI, Watertown, MA, USA) while carrying 23 kg that was evenly distributed in a weighted vest. Trials were conducted with and without a 3.2 kg replica F88-Austeyr rifle held in two hands across their body in a patrol carry position (weapon and no-weapon conditions) at four different speeds: self-selected (SS), 3.5 km□h$^{-1}$, 5.5 km□h$^{-1}$, and 6.5 km□h$^{-1}$. Self-selected speed was determined prior to experimental trials by having the participants walk on the treadmill and gradually increasing speed from 3.5 km□h$^{-1}$ until they reached their preferred speed. The speed was then set at 6.5 km□h$^{-1}$ and gradually decreased until a second preferred speed was reached. The final SS speed was taken as the mean of the two preferred speeds [27]. Participants were instructed to walk in the middle of the treadmill without holding the front bar of the treadmill. They were instructed to report any discomfort or fatigue that may have restricted them from finishing the trial, but no participant reported any issues. This study is part of a larger study exploring the effects of different physical constraints on soldiers' load carriage performance. Hence, due to the number and intensity of the trials, the conditions in this study were recorded over two sessions each one week apart in a counterbalanced design, this is; Session one (four trials) included trials at 5.5 km□h$^{-1}$ and SS speeds with and without weapon and Session two (four trials) included trials at 3.5 km□h$^{-1}$ and 6.5 km□h$^{-1}$ speeds with and without weapon.

*2.3. Instruments*

Four inertial sensors (APDM, Portland, OR, USA) sampling at 128 Hz were placed on the sternum (manubrium), sacrum, and feet (dorsal) using double sided tape and additional velcro straps for the feet sensors (Figure 1). Data were recorded and then exported for further analysis using APDM Moveo Explorer software.

*2.4. Data Analysis*

Gait stability was assessed using the short-term LDE (Lyapunov) using 3D linear accelerations. The LDE measures the ability of a person to deal with step-to-step perturbations, where higher LDE values (greater divergence) indicate lower stability [21]. Data from the foot sensors were used to determine heel contacts and extract sacrum and trunk acceleration time-series for the first 200 consecutive strides. Extracted data were normalised to 100 data points x 200 strides (20000 samples). 3D-LDEs for both sensors were calculated using a 9D state-space (3 x 3D delayed copies) with a time delay (t) of t = 6 for the $LDE_{SAC}$ and t = 10 for the $LDE_{STR}$. Time delays for each sensor were calculated as the median value across all trials and directions using the average mutual information algorithm. 3D short-term LDEs were calculated using the Rosenstein's algorithm[28] over the recommended 0-0.5 stride interval [29].

The secondary analysis in this study involved calculating LDE for each acceleration direction. The median embedding dimension (*m*) and delay (t) for each direction and for each sensor were used across all trials to calculate the sacrum and sternum vertical (VT), mediolateral (ML) and anteroposterior (AP) LDEs (Table 1). The rest of the steps for the LDE calculation were the same as for the 3D LDE. All calculations were performed in Matlab 2020b (Natick, USA).

### *2.5. Statistics*

Linear mixed model analyses were conducted to determine the main effects and interaction of weapon handling and speed for each of the sensors LDE (3D, VT, ML and AP) with a random effect of participant. Model residuals were assessed for normality using a Shapiro-Wilk tests and visually inspecting Q-Q plots. *Post-hoc* analyses with Bonferroni corrections were conducted to determine differences between speed conditions. For all analyses significance was set at $p < .05$. All analyses were conducted in Jamovi v.2.0.0.

## 3. Results

All participants were able to complete all trials, however, due to technical difficulties, data for some no weapon trials (2.2% of trials) were missing (two trials at 5.5 km□h$^{-1}$ and one at 6.5 km□h$^{-1}$; Supplementary Table 1). Participant's self-selected walking speed was on average 5.02 ± 0.23 km□h$^{-1}$ (females: 5.04 ± 0.30 km□h$^{-1}$ [4.87 – 5.21, 95% CI range]; males: 5.02 ± 0.22 km□h$^{-1}$ [4.81 – 5.24, 95% CI range]). Descriptive statistics for 3D LDE measures are presented in Table 2 and Figures 2 and 3. There were no interactions between speed and weapon handling on $LDE_{SAC}$ nor $LDE_{STR}$ ($p > .05$). We found a significant effect of speed on 3D $LDE_{SAC}$ ($F = 5.681$, $p = .001$) but no effect of weapon handling (F = .318, $p = .546$). *Post-hoc* tests showed a significantly larger $LDE_{SAC}$ at 3.5 km□h$^{-1}$ when compared with 5.5 km□h$^{-1}$ ($p = .003$) and SS ($p = .004$). There was a significant effect of both weapon handling ($F = 24.963$, $p < .001$) and speed

($F$ = 106.057, $p$ < .001) on $LDE_{STR}$. *Post-hoc* tests showed that $LDE_{STR}$ was significantly lower (more stable) when handling a weapon when compared with no weapon, and as walking speed increased ($p$ < .001), except between SS and 5.5 km□h$^{-1}$ ($p$ = .338) and 6.5 km□h$^{-1}$ and 5.5 km□h$^{-1}$ ($p$ = .052).

There were no interactions between speed and weapon handling on $LDE_{SAC}$ nor $LDE_{STR}$ ($p$ > .05) when analysed per direction of acceleration. For $LDE_{SAC}$, a significant effect of weapon handling was only observed in the VT direction ($p$ = .0048). We found a significant effect of speed for the VT ($F$ = 8.296, $p$ < .001) and AP ($F$ = 15.960, $p$ < .001) directions. *Post-hoc* comparisons showed that differences in $LDE_{SAC}$ occurred between 3.5 km□h$^{-1}$ ($p$ < .001) and SS ($p$ = .045) compared to 6.5 km□h$^{-1}$ in the VT direction and between 3.5 km□h$^{-1}$ and all other speeds in the AP direction ($p$ < .001). For $LDE_{STR}$, significant effects of weapon handling and speed were found in all three directions ($p$ < .001). *Post-hoc* comparisons revealed that $LDE_{STR}$ was significantly lower (more stable) when handling a weapon when compared with no weapon, and differences for speed occurred between 3.5 km□h$^{-1}$ and all other speeds across all directions ($p$ < .001), between SS and 6.5 km□h$^{-1}$ in all directions ($p$ < .0451) and 5.5 km□h$^{-1}$ and 6.5 km□h$^{-1}$ in the ML and AP directions ($p$ < .016). A table with all post-hoc comparisons per sensor and per direction is provided in supplementary table 2.

## 4. Discussion

The primary aim of this study was to determine the effects of weapon handling and walking speed on overall (3D) gait stability in soldiers. Our secondary aim was to determine if stability in any direction was also affected by these changes in task constraints. We found that handling a weapon increased stability at the sternum (lower $LDE_{STR}$) but had no effect at the sacrum ($LDE_{SAC}$). Faster walking speeds, on the other hand, increased stability (lower LDE values) at both the sacrum ($LDE_{SAC}$) and sternum ($LDE_{STR}$). The main effect of speed on $LDE_{SAC}$ was mainly due to lower stability when walking at 3.5 km□h$^{-1}$ when compared with 5.5 km□h$^{-1}$, while $LDE_{STR}$ differences occurred between all speeds except SS (5.3 km□h$^{-1}$) and 5.5 km□h$^{-1}$.

A speed of 3.5 km□h$^{-1}$ is considered a slow-to-moderate marching speed in soldiers [30] and it was slower than the SS speed of our participants. Since stability was the lowest at this speed, walking at this speed on a flat uniform surface may be an intrinsically more unstable condition, regardless of the load and weapon carried [31]. In fact, walking slowly may be avoided when dealing with stability-threatening perturbations that may lead to a fall, for example, [32] even when slow walking may allow more time to find the best way to sort obstacles [14]. On the positive side, lower kinetic energy when walking slow may reduce the consequences of a fall [14].

On the other hand, time constraints to respond to internal/external perturbations during fast walking may indicate an increased cortical control, which is thought to prime the sensorimotor system for performing timely gait adjustments [33]. Hence, it is possible that there is an increasing involvement of voluntary control and cognitive resources as speed increases, which is reflected in the stability increases found with increasing walking speed. Although young adults' walking performance as speed increases does not seem to be affected by cognitive tasks [34], in the military context, cognitive resources may be in greater demand and be affected by increasing walking speed.

$LDE_{SAC}$ were relatively similar across SS, 5.5 km□h$^{-1}$, and 6.5 km□h$^{-1}$, whereas $LDE_{STR}$ decreased with increasing speed. This effect discrepancy between sacrum and sternum LDE may be due to the proximity of the former sensor to the CoM, which despite its greater excursion when carrying a weapon [11] or loaded, is more tightly controlled through increased stiffness[16], for example, in order to maintain stability[35]. The latter also extends to explain similar findings in the LDE directional analyses and may further supports the notion of the CoM as the main controlled variable in human motion[36]. A study exploring the effects on stability (LDE) of a backpack carrying an in- or out-of-phase inverted pendulum found a reduced motion of the CoM and increased stability in both conditions compared to a fixed pendulum [35]. These results further emphasize the direct control exerted over the CoM to maintain stability and are in line with our interpretation of the need to prioritise CoM control to maintain stability when dealing with external constraints and perturbations [35].

Increased stability at the trunk ($LDE_{STR}$) may be due to movement restrictions imposed by the weighted vest, which may be further restricted when handling a weapon. Weapon carriage displaces the CoM forward [10] while simultaneously constraining the arms' natural swing, leading to increased pelvis-trunk coordination, coupled with increased variability in the transverse plane [9]. We found that restricting arm swing through carrying a weapon increased stability, however, this effect has also been found when, conversely, actively (exaggerated) swinging the arms [12,13]. Taken together, this may indicate a non-linear effect of arm swing on stability of the trunk when walking[13,15,37]. We found that stability increased with speed even when with no weapon, corroborating findings from previous studies in which stability was associated with speed and arm swing increases [13,15,37].

The combination of load carriage and weapon handling constraints may also impose a more direct control (voluntary) of trunk musculature to increase stability[38], particularly when walking at the fastest speeds[33]. This mechanism may allow a tighter control of balance in potentially more unstable conditions where similar magnitude perturbations may have a larger effect[15]. Interestingly, Walsh et al. (2021) found that stability of trunk-muscle activation was lower when carrying an 11-kg webbing when compared with no load[22]. It has been proposed previously that restricting arm swing does not decrease stability and that unrestricted arm swing may be helpful in recovering after a perturbation [15]. Our finding of greater trunk stability when handling a weapon may, therefore, make it difficult for soldiers to stabilize after a perturbation due to restrictions of the arms' movements. However, it is also possible that trunk stabilization may be a proactive measure in the case of having to cope with potentially destabilizing external forces.

For the sacrum LDEs, speed affected stability in a similar fashion as in previous studies in which VT LDE increased and AP LDE decreased as speed increased [14,26]. However, these studies used a broader range of speeds and data type to calculate the LDE, for example, Bruijn et al (2009) used thorax marker velocity (T6 level) and Punt et al (2015) used lower back velocity. When comparing to Bruijn et al's results we found an opposite direction of the speed effect on the sternum's VT LDE, this is; stability increased as speed increased. This may be explained by the fact that our participants were carrying a 23-kg vest that may have forced a greater vertical control of the CoM when loaded [11,35]. This increased vertical displacement control is likely the result of a greater activation of trunk musculature [22] and not stepping behaviour adjustments [35] and may also be associated to an increased cost of walking as weapon carriage limits arms swing [39]. The opposite effect of speed on the VT LDE at the sacrum and sternum may indicate a compensatory relationship to deal with speed and weapon handling, nonetheless, this relationship may not be linear and is yet to be explored.

For the ML direction, we only found speed and weapon handling effects for the ML $LDE_{STR}$ showing greater stability when carrying a weapon and increased stability (lower LDE values) as speed increased. The latter sped effect has been previously reported [31], yet differs from an inverted U pattern in LDE values with peak at about 4.6 $km \cdot h^{-1}$ (2.2 $km \cdot h^{-1}$ to 6.2 $km \cdot h^{-1}$ speed range) [26]. Our results are in line with previous studies suggesting the need for greater ML CoM control as speed increases [40], and to handle the potential destabilizing effect of increasing ML impulse when carrying a weapon [11].

Overall, with and without weapon handling our $LDE_{STR}$ directional results follow the same trend as in Stenum et al (2014) when utilizing the same LDE calculation methodology and sensor location (method C; normalization of n stride data *100 and LDE over the 0-0.5 stride range) [31]. On the other hand, our per-direction results for the $LDE_{SAC}$ have a similar trend to the results reported by Punt et al. (2015), who used data collected at a similar location (lower back) and analyzed with the same methodology, when comparing similar speed ranges. Interestingly, previous studies exploring the effect of speed on stability have employed a broader range of speeds (approximately 1 to 7 km□h$^{-1}$) with the steepest LDE changes at speeds <4 km□h$^{-1}$ [14,26]. However, walking speeds slower than 3.5 km□h$^{-1}$ are not commonly employed during military duties and were not addressed in the present study.

The participants in this study were experienced soldiers with no injuries and from whom LDE results can be used as the first step in determining LDE reference values for common military duties. Although the 3D LDE can provide an overall view of the effects of different constraint during military marching on stability, the use of directional LDE values may be useful to determine if such constraints may elicit specific balance control responses. This may also help in identifying tailored training methods or strategies that can better address the demands of military marching and prevent musculoskeletal injuries. For example, asymmetrical frontal-plane loading, or hip-abductor fatigue/injury may lead to more specific increases of ML LDE (lower ML stability), which can be reduced by better loading arrangement or physiotherapy/training interventions of hip musculature. Further research should be conducted to determine if the LDE (3D or any direction) can be used as a sensitive biomarker of musculoskeletal injuries during training as well as stablishing reference values for clinical decision making.

*Limitations*

Our study explored weapon handling and speed effects on gait stability while walking on a treadmill, which is known to affect most gait measures, including the LDE, when compared to overground walking[41]. Further studies exploring stability in more representative environments are warranted. To note, however, treadmill walking may be more suitable when determining rehabilitation effects after an injury in the military as it is a safer and controllable environment.

We also used a single load (23-kg weighted vest), which is close to the minimum soldiers carry during combat. However, the effects of weapon handling and speed when carrying larger loads and in different locations (i.e. backpack) is yet to be explored [1]. There are different methods used to calculate LDE that may yield different results when exploring the effects of weapon handling and speed [31]. However, the methods used in the present study are widely accepted, hence, we are confident of the results [21,31,42]. Non-linear measures of walking, such as the LDE are at least as important as linear features and may offer a better understanding of the systems resilience to perturbations that may lead to injuries[20]. Finally, LDE is not the only measure used to explore the effects of external constraints in gait stability[43], but its use is supported by its construct, predictive, and convergent validity[21].

## 5. Conclusions

This study found that local dynamic stability measured at the sacrum was not affected by weapon handling and speed reduced stability only when walking slow. Contrary to what it may be thought, weapon handling increased gait stability measured at the sternum, however, this may be a strategy to support maintaining stability at the sacrum in conditions that are more vulnerable to perturbations. Our findings may help trainers and clinicians to identify soldiers at a greater risk of injuries when unable to maintain adequate gait stability during military tasks.

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## Declaration of interest

The authors declare no conflict of interest.

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**Figure 1.** Representative data collection set-up with participant walking on an instrumented treadmill while carrying a 23 kg weighted vest and a 3.2 kg replica F88-Austeyr rifle. To note, in this study we only utilized inertial sensors at the sacrum and sternum for the LDE calculations, whilst the foot sensors were used for heel strike identification.

**Figure 2.** Boxplots presenting local divergence exponent (LDE) values for the sacrum sensor in each direction (VT, ML and AP) and 3D for both no weapon (blue) and weapon (yellow) conditions across all speeds. * Post-hoc significant speed differences.

**Figure 3.** Boxplots presenting local divergence exponent (LDE) values for the sternum sensor in each direction (VT, ML and AP) and 3D for both no weapon (blue) and weapon (yellow) conditions across all speeds. * Post-hoc significant speed differences.

**Dear Xingda Qu**
**Editor Ergonomics Journal**

Once again, we would like to thank the reviewers for their insightful comments and suggestions. We are also very thankful for the positive comments from reviewers #2 and #3, who are happy with our revised document. We have carefully read the feedback, particularly from Reviewer 1, to which we have replied on a point-by-point basis as follows:

**Reviewer 1 (R1)**

Authors have addressed reviewers' comments with more detail descriptions on data collection and analysis. New references have been added to justify the study design and support study findings. While the overall readability has been improved by revision, the manuscript still suffers from following limitations or concerns.

**Q1.** As authors mentioned, the evaluation of dynamic stability during walking has been a common method to assess individual's walking balance capacity or fall risks. An issue regarding the evaluation of LDE in this study is the lack of evidence that justifies the need for balance assessment. Authors have listed a reference that shows the high incidence of musculoskeletal injuries of soldiers during load carriage or marching training. However, the reference does not show any evidence that such injuries resulted from falls or impaired walking balance while walking with load. That is, the need for the evaluation of dynamic walking balance for this specific population and the specific operations (walking with loads) was not justified.

**To directly address the issue of falls, injuries, load carriage, walking and falls associations, the following text (and citations) has been added to Paragraph 1 in the Introduction:**
***"For example, walking and training while carrying loads have been identified as modifiable and preventable non-battle causes of fall-related injuries[1-3]"***

**Q2.** One of major issues that were mentioned by multiple reviewers is the lack of a clear association between study findings and injury prevention. The new paragraph in Discussion does not explain well the association.

a) How can we use the LDE values of the current study as reference values in future studies? Specifically, considering the small sample size, it is not convincing that the data of the current study would serve as normative data or reference values.

**We have stated that "…LDE results can be used as the first step in determining LDE reference values for common military duties". We do not intend, considering our sample size and composition, to provide a definitive value for the LDE. In the military, as well as in the neurology field, non-linear metrics rely on a few parameters including number of embedded dimensions and time delay, which allow to properly capture walking dynamic stability. Consensus regarding, at least, these parameters needs to be achieved before suggesting normative values. However, under the current (fully described and justified) parameters we are able to provide a first picture about how speed and load constraints affect dynamic walking.**

b) In addition, how can we use the study findings to identify tailored training methods?
**To make this point more explicit, the following text has been added:**
**"For example, asymmetrical frontal-plane loading or hip-abductor fatigue/injury may lead to more specific increases of ML LDE (lower ML stability), which can be reduced by better loading arrangement or physiotherapy/training interventions of hip musculature."**

c) Is it always better to lower the LDE values to lower risks for musculoskeletal injuries?
**The LDE "quantifies the average logarithmic rate of divergence of a system after a small perturbation". Hence, in a cyclic task such as walking (or running) reducing the LDE values is an indication of a greater ability to deal with ongoing perturbations arising at every step, which is**

**the most likely reflection of indemnity of the neuromuscular system and reduced risk of MSK injuries.**
**It is noteworthy, however, that the current study design does not intend to answer this question, which can be addressed by a follow-up or a pre-post intervention study. Nevertheless, as a reference a few studies in clinical populations have shown that the LDE improves after rehabilitation and is associated to prospective falls [4,5].**

**Q3.** The results of LDE evaluation simply indicate the size of directional divergence of the CoM during cyclic walking. Fall risk evaluation should be done not only by the CoM tracking but also CoP monitoring. Individuals may walk with wider CoP base when walking slowly, and it may result in the greater LDE values.

**Regarding the first part of the question, we are NOT intending to create a fall risk measure, although we acknowledge that due to the LDE's construct validity in other populations it may well serve as one. Biomechanical measures of the CoM or CoP have long shown to be reflective of walking stability, however, they are constrained to lab settings. Perhaps a further validation step, in the military context, would be to determine the associations between mechanical and non-linear dynamic measures, however, that's beyond the scope of this study.**

**In the second part of the question, the reviewer hypothesises about the effect of mechanical adaptations on dynamic behaviour. To our knowledge, in the military context, there is no evidence of such association. However, from clinical studies ([6-8], as examples) it has been shown that the LDE is affected even in the absence of spatiotemporal measures differences, which indicates its potential as a more sensitive measure of dynamic behaviour.**

**<u>Reviewer 2 (R2)</u>**

The authors have answered my comments thoughtfully. I have no further comments.

**Thank you**

**<u>Reviewer 3 (R3)</u>**

Congratulations to the authors for their revisions. This reviewer has a few suggestions which will hopefully improve the readership of the manuscript:

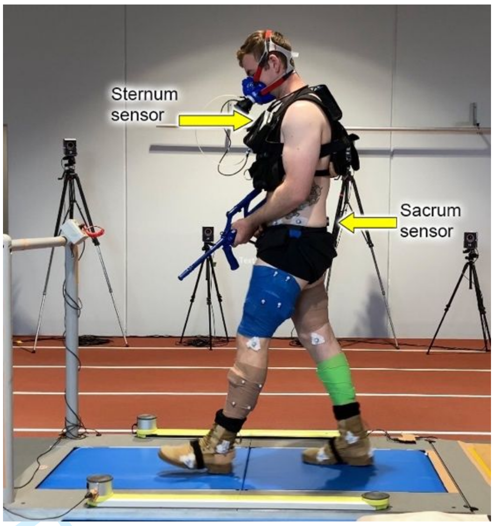
**Thanks for the positive comment**

**Q1.** Results section, recommend adding a sentence after first sentence which describes that data for one participant was missing during the 6.5 km*h-1 condition with no weapon, and two values were missing during 5.5 km*h-1 speed with no weapon condition. If this can be added the reader will not have to refer to the supplemental Table 1.

**To address this comment, the text at the beginning of the Results section has been modified and now read as follows:**
***"…..no weapon trials (2.2% of trials) were missing (two trials at 5.5 km$\cdot$h$^{-1}$ and one at 6.5 km$\cdot$h$^{-1}$; Supplementary Table 1)."***

**Q2.** Figure 1. Upon reflection, since these data were analyzed from a subset of sensors that were on the participants, this reviewer recommends the authors somehow indicate the approximate location of the IMUs that were used for current study (using the current photo).

**Please refer to figure 1 (copy below), which has been modified to address this comment.**

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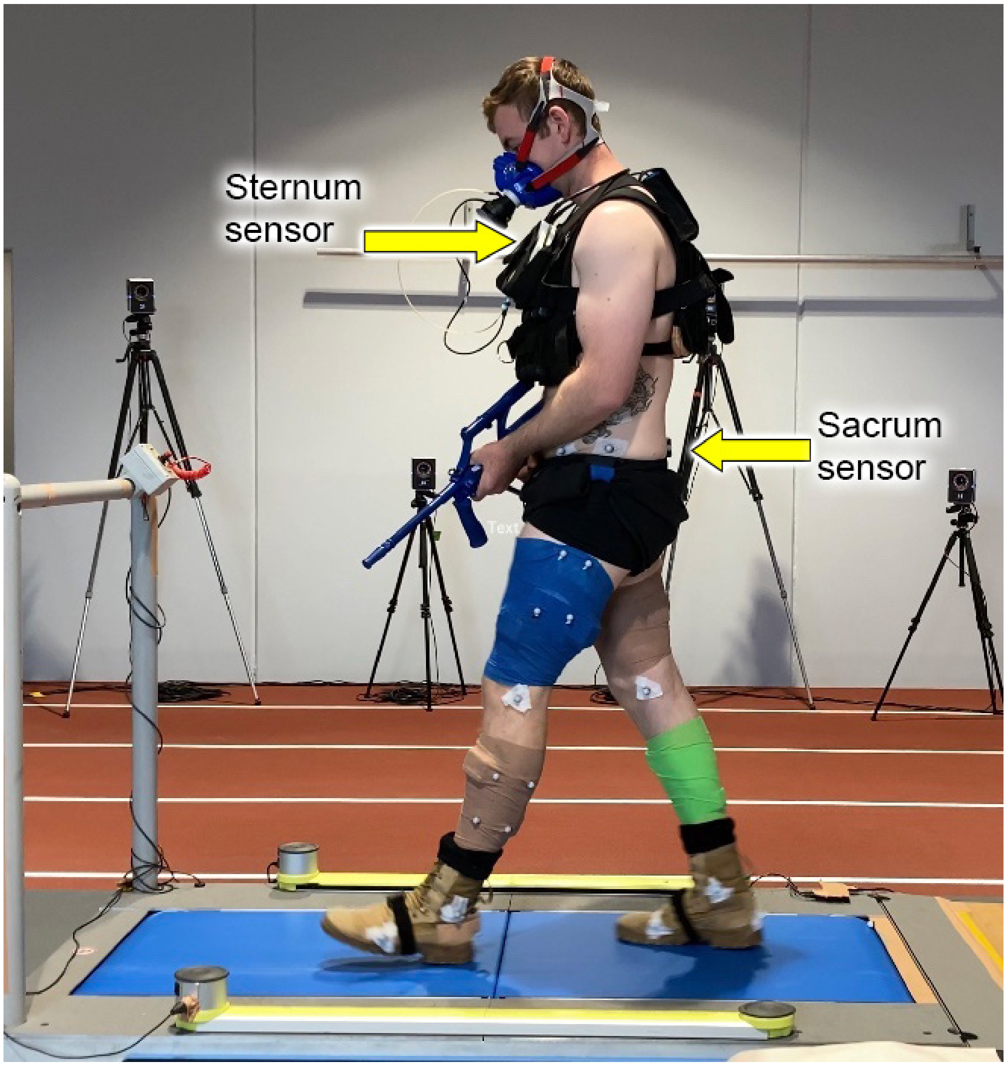
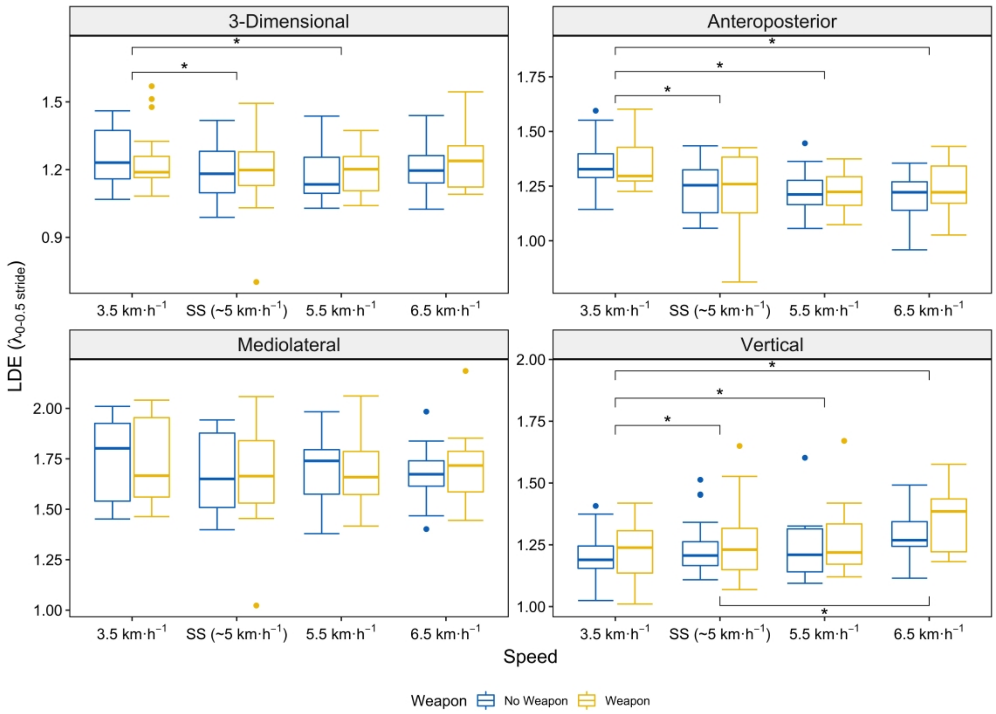


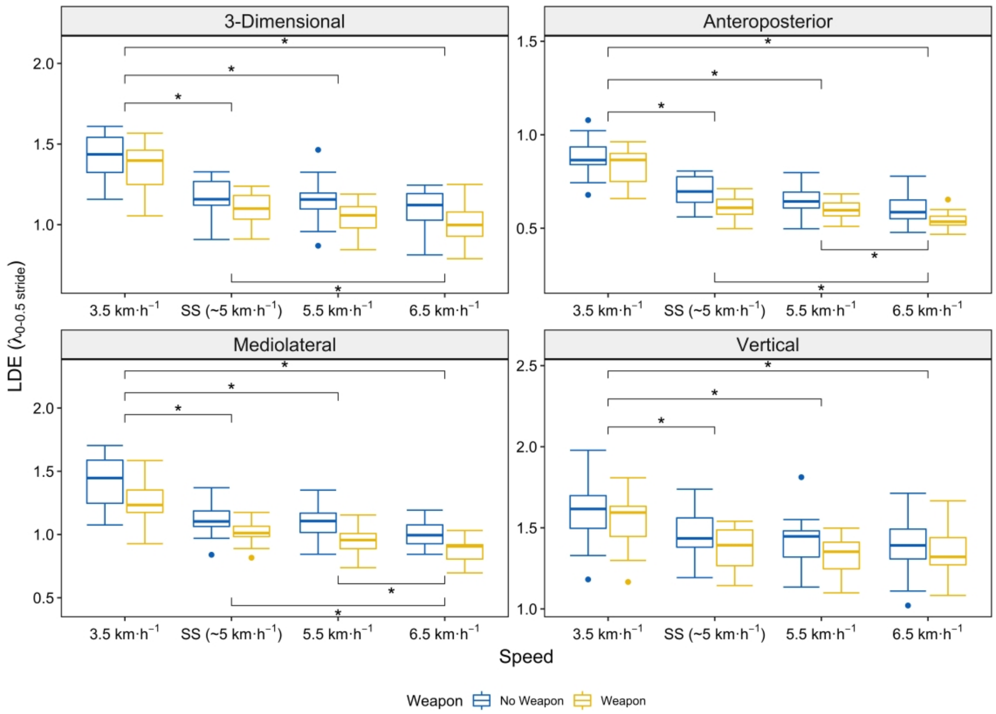

Figure 1. Representative data collection set-up with participant walking on an instrumented treadmill while carrying a 23 kg weighted vest and a 3.2 kg replica F88-Austeyr rifle. To note, in this study we only utilized inertial sensors at the sacrum and sternum for the LDE calculations, whilst the foot sensors were used for heel strike identification.

232x246mm (72 x 72 DPI)




Boxplots presenting local divergence exponent (LDE) values for the sacrum sensor in each direction (VT, ML and AP) and 3D for both no weapon (blue) and weapon (yellow) conditions across all speeds. * Post-hoc significant speed differences.

2469x1795mm (72 x 72 DPI)




Boxplots presenting local divergence exponent (LDE) values for the sternum sensor in each direction (VT, ML and AP) and 3D for both no weapon (blue) and weapon (yellow) conditions across all speeds. * Post-hoc significant speed differences.

2469x1795mm (72 x 72 DPI)

**Table 1.** Median embedding dimensions (*m)* and delay (t) values for each sensor and plane.

| | Anteroposterior (AP) | | Mediolateral (ML) | | Vertical (VT) | |
|---|---|---|---|---|---|---|
| | ***m*** | **t** | ***m*** | **t** | ***m*** | **t** |
| **Sternum** | 6 | 12 | 7 | 10 | 6 | 8 |
| **Sacrum** | 7 | 6 | 7 | 4 | 6 | 7 |

**Table 2.** Local divergent exponent (LDE) descriptive statistics of the 3D sacrum ($LDE_{SAC}$) and sternum ($LDE_{STR}$). Values are presented as mean (standard deviation (SD)) and 95% confidence interval (CI). *significant effect of speed ($p < .05$), #significant effect of weapon handling ($p < .05$).

| | | No weapon | | | | Weapon handling | | | |
|---|---|---|---|---|---|---|---|---|---|
| | | 3.5 km h$^{-1}$ | SS (~5.3 km h$^{-1}$) | 5.5 km h$^{-1}$ | 6.5 km h$^{-1}$ | 3.5 km h$^{-1}$ | SS (~5.3 km h$^{-1}$) | 5.5 km h$^{-1}$ | 6.5 km h$^{-1}$ |
| $LDE_{SAC}$ *# | Mean (SD) | 1.26 (0.12) | 1.19 (0.13) | 1.18 (0.12) | 1.20 (0.1) | 1.25 (0.14) | 1.19 (0.18) | 1.19 (0.11) | 1.24 (0.12) |
| | 95%CI | [1.20, 1.31] | [1.13, 1.25] | [1.12, 1.25] | [1.15, 1.25] | [1.18, 1.32] | [1.11, 1.28] | [1.14, 1.24] | [1.18, 1.29] |
| $LDE_{STR}$ # | Mean (SD) | 1.43 (0.14) | 1.18 (0.11) | 1.15 (0.14) | 1.09 (0.13) | 1.37 (0.15) | 1.11 (0.10) | 1.05 (0.10) | 1.01 (0.12) |
| | 95%CI | [1.36, 1.49] | [1.13, 1.23] | [1.07, 1.22] | [1.03, 1.15] | [1.30, 1.44] | [1.06, 1.16] | [1.00, 1.09] | [0.95, 1.06] |

**Supplementary Table 1.** Ticks indicate recorded and processed trials whereas crosses indicate those trials that were not processed due to technical issues during the recording.

| | Speed 3.5 kmh$^{-1}$ | | Self-selected (SS) | | Speed 5.5 kmh$^{-1}$ | | Speed 6.5 kmh$^{-1}$ | |
|---|---|---|---|---|---|---|---|---|
| **Subject** | **Weapon** | **No Weapon** | **Weapon** | **No Weapon** | **Weapon** | **No Weapon** | **Weapon** | **No Weapon** |
| **1** | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ |
| **2** | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ |
| **3** | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ |
| **4** | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ |
| **5** | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ |
| **6** | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ |
| **7** | ✓ | ✓ | ✓ | ✓ | ✓ | ✗ | ✓ | ✓ |
| **8** | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ |
| **9** | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✗ |
| **10** | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ |
| **11** | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ |
| **12** | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ |
| **13** | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ |
| **14** | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ |
| **15** | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ |
| **16** | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ | ✓ |
| **17** | ✓ | ✓ | ✓ | ✓ | ✓ | ✗ | ✓ | ✓ |

**Supplementary Table 2.** Descriptive statistics and mixed model results. Post-hoc comparisons for speed are presented on the right side of the table. All significant effects are highlighted in bold (*p*<.05).

| | | | No weapon | | | | Weapon handling | | | | Weapon* Speed | | Weapon | | Speed | | Post-hoc speed comparisons (Bonferroni corrected) | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | | | 3.5 $kmh^{-1}$ | SS - 5 $kmh^{-1}$ | 5.5 $kmh^{-1}$ | 6.5 $kmh^{-1}$ | 3.5 $kmh^{-1}$ | SS - 5 $kmh^{-1}$ | 5.5 $kmh^{-1}$ | 6.5 $kmh^{-1}$ | *F* | *p* | *F* | *p* | *F* | *p* | 3.5 vs 5.5 | 3.5 vs 6.5 | 3.5 vs SS | SS vs 5.5 | SS vs 6.5 | 5.5 vs 6.5 |
| 3 dimensional (3D) | Sacrum | mean | 1.26 | 1.19 | 1.18 | 1.20 | 1.25 | 1.19 | 1.19 | 1.24 | 0.55 | 0.652 | 0.32 | 0.574 | **5.7** | **0.001** | **0.003** | 0.372 | **0.004** | 1.000 | 0.715 | 0.555 |
| | | sd | 0.12 | 0.13 | 0.12 | 0.10 | 0.14 | 0.18 | 0.11 | 0.12 | | | | | | | | | | | | |
| | | CI upper | 1.20 | 1.13 | 1.12 | 1.15 | 1.18 | 1.11 | 1.14 | 1.18 | | | | | | | | | | | | |
| | | CI lower | 1.31 | 1.25 | 1.25 | 1.25 | 1.32 | 1.28 | 1.24 | 1.29 | | | | | | | | | | | | |
| | Sternum | mean | 1.43 | 1.18 | 1.15 | 1.09 | 1.37 | 1.11 | 1.05 | 1.01 | 0.59 | 0.624 | **24.96** | **< .001** | **106.1** | **< .001** | **< .001** | **< .001** | **< .001** | 0.338 | **< .001** | 0.052 |
| | | sd | 0.14 | 0.11 | 0.14 | 0.13 | 0.15 | 0.10 | 0.10 | 0.12 | | | | | | | | | | | | |
| | | CI upper | 1.36 | 1.13 | 1.07 | 1.03 | 1.30 | 1.06 | 1.00 | 0.95 | | | | | | | | | | | | |
| | | CI lower | 1.49 | 1.23 | 1.22 | 1.15 | 1.44 | 1.16 | 1.09 | 1.06 | | | | | | | | | | | | |
| Vertical (VT) | Sacrum | mean | 1.20 | 1.25 | 1.24 | 1.29 | 1.21 | 1.26 | 1.27 | 1.35 | 0.72 | 0.543 | **3.99** | **0.048** | 8.3 | < .001 | **0.156** | **< .001** | **0.149** | 1.000 | **0.045** | 0.055 |
| | | sd | 0.10 | 0.12 | 0.13 | 0.11 | 0.12 | 0.15 | 0.14 | 0.12 | | | | | | | | | | | | |
| | | CI upper | 1.15 | 1.19 | 1.17 | 1.23 | 1.15 | 1.19 | 1.20 | 1.29 | | | | | | | | | | | | |
| | | CI lower | 1.24 | 1.31 | 1.30 | 1.34 | 1.27 | 1.33 | 1.33 | 1.41 | | | | | | | | | | | | |
| | Sternum | mean | 1.60 | 1.45 | 1.41 | 1.39 | 1.54 | 1.37 | 1.33 | 1.35 | 0.49 | 0.687 | **11.19** | **0.001** | **21.5** | **< .001** | **< .001** | **< .001** | **< .001** | 1.000 | 0.527 | 1.000 |
| | | sd | 0.20 | 0.15 | 0.16 | 0.20 | 0.17 | 0.13 | 0.12 | 0.17 | | | | | | | | | | | | |
| | | CI upper | 1.51 | 1.38 | 1.32 | 1.29 | 1.46 | 1.31 | 1.27 | 1.27 | | | | | | | | | | | | |
| | | CI lower | 1.70 | 1.52 | 1.49 | 1.49 | 1.62 | 1.43 | 1.39 | 1.43 | | | | | | | | | | | | |
| Mediolateral (ML) | Sacrum | mean | 1.72 | 1.68 | 1.69 | 1.66 | 1.75 | 1.67 | 1.68 | 1.70 | 0.42 | 0.742 | 0.42 | 0.520 | 2.0 | 0.121 | 0.442 | 0.310 | 0.215 | 1.000 | 1.000 | 1.000 |
| | | sd | 0.21 | 0.19 | 0.16 | 0.15 | 0.21 | 0.25 | 0.16 | 0.18 | | | | | | | | | | | | |
| | | CI upper | 1.63 | 1.59 | 1.61 | 1.59 | 1.65 | 1.55 | 1.60 | 1.62 | | | | | | | | | | | | |
| | | CI lower | 1.82 | 1.77 | 1.77 | 1.73 | 1.85 | 1.79 | 1.75 | 1.79 | | | | | | | | | | | | |
| | Sternum | mean | 1.42 | 1.13 | 1.10 | 1.01 | 1.26 | 1.02 | 0.95 | 0.88 | 0.42 | 0.742 | **52.20** | **< .001** | **85.8** | **< .001** | **< .001** | **< .001** | **< .001** | 0.359 | **< .001** | **0.014** |
| | | sd | 0.20 | 0.13 | 0.13 | 0.11 | 0.16 | 0.09 | 0.11 | 0.10 | | | | | | | | | | | | |
| | | CI upper | 1.33 | 1.07 | 1.03 | 0.95 | 1.18 | 0.97 | 0.90 | 0.83 | | | | | | | | | | | | |
| | | CI lower | 1.51 | 1.19 | 1.17 | 1.06 | 1.33 | 1.06 | 1.00 | 0.93 | | | | | | | | | | | | |
| Anteriorposterior (AP) | Sacrum | mean | 1.34 | 1.24 | 1.23 | 1.20 | 1.36 | 1.25 | 1.22 | 1.23 | 0.27 | 0.846 | 0.62 | 0.433 | **16.0** | **< .001** | **< .001** | **< .001** | **< .001** | 1.000 | 1.000 | 1.000 |
| | | sd | 0.12 | 0.12 | 0.10 | 0.10 | 0.12 | 0.16 | 0.08 | 0.12 | | | | | | | | | | | | |
| | | CI upper | 1.29 | 1.18 | 1.18 | 1.15 | 1.30 | 1.17 | 1.18 | 1.17 | | | | | | | | | | | | |
| | | CI lower | 1.40 | 1.29 | 1.28 | 1.25 | 1.42 | 1.32 | 1.26 | 1.29 | | | | | | | | | | | | |
| | Sternum | mean | 0.88 | 0.70 | 0.65 | 0.60 | 0.83 | 0.61 | 0.59 | 0.54 | 0.48 | 0.700 | **33.21** | **< .001** | **143.4** | **< .001** | **< .001** | **< .001** | **< .001** | 0.115 | **< .001** | **0.016** |
| | | sd | 0.10 | 0.08 | 0.08 | 0.09 | 0.10 | 0.06 | 0.05 | 0.04 | | | | | | | | | | | | |
| | | CI upper | 0.84 | 0.66 | 0.61 | 0.56 | 0.78 | 0.59 | 0.57 | 0.52 | | | | | | | | | | | | |
| | | CI lower | 0.93 | 0.73 | 0.69 | 0.65 | 0.88 | 0.64 | 0.62 | 0.57 | | | | | | | | | | | | |