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A generalisable methodology for stability assessment of walking aid users

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ABSTRACT

To assist balance and mobility, older adults are often prescribed walking aids. Nevertheless, surprisingly their use has been associated with increased falls-risk. To address this finding we first need to characterise a person's stability while using a walking aid. Therefore, we present a generalisable method for the assessment of stability of walking frame (WF) users. Our method, for the first time, considers user and device as a combined system.

We define the combined centre of pressure (CoP_{system}) of user and WF to be the point through which the resultant ground reaction force for all feet of both the WF and user acts if the resultant moment acts only around an axis perpendicular to the ground plane.

We also define the combined base of support (BoS_{system}) to be the convex polygon formed by the boundaries of the anatomical and WF feet in contact with the ground and interconnecting lines between them. To measure these parameters we have developed an instrumented WF with a load cell in each foot which we use together with pressure-sensing insoles and a camera system, the latter providing the relative position of the WF and anatomical feet. Software uses the resulting data to calculate the stability margin of the combined system, defined as the distance between CoP_{system} and the nearest edge of BoS_{system}. Our software also calculates the weight supported through the frame and when each foot (of user and/or frame) is on the floor. Finally, we present experimental work demonstrating the value of our approach.

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1. Introduction

Falls in older adults are a major global health problem as more than 30% of community-dwelling people aged 65 and over fall every year [1], consequences of which range from reduced activity and fear of falling to injuries and death [2]. Moreover, falls are also a matter of great concern for society as a whole: in 2013, for instance, it was estimated that falls cost the UK government over £2.3 billion [3]. Older frail people with an unstable gait are often advised by their clinician to use walking aids, which are designed to help them maintain their balance through an increase in the effective base of support area, and through provision of structural support and haptic sensory information [4,5]. Indeed, walking aids are used by 29–49% of older people [6]. However, paradoxically, use of walking aids (versus non-use) has been associated with a 2-fold to 3-fold increase in risk of falling [7]. There are a number of possible explanations for this finding: one is that walking aids are prescribed to the most frail part of the population who, when falls occur, are most likely to suffer injury and, hence, appear in the statistics; another is that prescription of a walking aid increases the period spent upright or mobile and, hence, reduces time spent in a safer sitting or lying posture. However, in studies by Mann et al. [8] and Skymne et al. [9], 60% of walking frame (WF) users reported problems with using their frame and quotations from users included "(the frame was) difficult and/or dangerous to use" and "…could it (the walking frame) overturn when used; was it really stable?". Such concerns suggest that another possible explanation and the motivation for this work, is that incorrect device usage, as a result of inappropriate device selection and/or training, may be contributing to instability and falls in WF users.

Surprisingly, despite the large number of walking aid users amongst the older population, there are no objective methods,

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generalisable methods for assessing their stability. Previous work to date has often focused on the kinematics/kinetics of the user only, presuming that the more the gait pattern resembles that of a healthy subject, the more stable the user is [9–14]. Such an approach ignores any direct effects of the walking frame on the user's stability, which is clearly incorrect [15]. Others focused on the device alone [16,17]: Pardo et al., for instance, developed an instrumented walking frame to detect lift-off/touch-down events of the frame itself and to calculate device loading and device Centre of Pressure (CoP) [18]. They inferred stability by assuming that, if the device CoP approaches the boundaries of its Base of Support (BoS) and, therefore, if the WF becomes unstable, then, the higher the loading on the device, the higher the risk of falling. To quantify stability, they derived the Walker Tipping Index (which gives an indication of how close the device is from tipping) from the horizontal and vertical forces applied to the walking frame, and then normalised such index to the percentage of body weight transferred onto the device [19]. However, the walking frame and user are mechanically coupled and determining when tipping is imminent based on a measure of either the mechanics of the user alone, or their frame alone, is incorrect. For example, when the WF is being lifted, initially only two WF feet remain in contact with the ground, and the frame CoP lies on the boundary of the frame BoS, which is reduced to the line connecting the two grounded feet. A measure that only considers the WF would interpret this scenario as being unstable, whilst this is, in fact, a natural part of WF use. Therefore, although it is true that tipping of the walking frame might mean that the user has fallen, it is more likely to indicate that the user is beginning to lift the walker. Similarly, measures based on the frame alone cannot inform on stability when the device is fully airborne which is likely to constitute a particularly challenging situation to the user. Conversely, when the user is relying on the walker, it is likely that the CoP of the user alone is under the user's toes and, hence, very close to the edge of the user's BoS; however, this does not mean that the user is unstable, rather that they are leaning on the device. Only one study to date collected data on both user and their device (a rollator) [20]. Whilst their approach is praiseworthy, stability of the overall system (defined as person and walking aid) was not adequately addressed because the mechanics of the user and their walking aid were treated separately and stability was evaluated on the basis of reliance on the device and excursions of the device centre of pressure.

The whole system, comprising user and frame may be considered to be a configurable multi-legged device, similar to a multi-legged walking robot. Methods for the calculation of stability of multi-legged robots based on the CoP kinematics are well established [21–24] and are directly applicable to this problem. Yet stability methods from the robotics literature have not been previously reported in the context of walking aid usage. Considering user and device as a combined system has the advantage of allowing for the correct assessment of stability under all user-frame configurations, including when the WF is airborne, which may be particularly critical.

This paper proposes an objective and generalisable method for the assessment of stability of walking aid users, based on methods from the robotics literature. Given that there are more walker users than users of crutches [25] and since seven times as many injuries are associated with walkers compared with walking sticks [26], we here introduce our method for the assessment of stability of walker usage, specifically for a walking frame without wheels (a pick-up walker). We demonstrate the application of the methods for walking in a standardized home environment, the University of Salford Activities of Daily Living (ADL) flat.

2. Methods

2.1. Stability of the system (user and walking frame)

The novel methods proposed here consider the user and their four legged walking frame (WF) as a combined system. We define the combined centre of pressure (CoP_{system}) of user and WF to be the point through which the resultant ground reaction force for all feet of both the WF and user acts if the resultant moment acts only around an axis perpendicular to the ground plane.

The instantaneous position of the combined CoP is calculated as follows:

$$\boldsymbol{COP_x} = \frac{\sum_{i=1}^{n}(\boldsymbol{Fv_i x_i})}{\sum_{i=1}^{n}\boldsymbol{Fv_i}} \qquad \boldsymbol{COP_y} = \frac{\sum_{i=1}^{n}(\boldsymbol{Fv_i y_i})}{\sum_{i=1}^{n}\boldsymbol{Fv_i}} \tag{1}$$

where:

- $COP_{x,y}$ are the coordinates of the CoP in the mediolateral and anteroposterior direction, respectively;
- Fv_i is the vertical load on the ith supporting foot (either anatomical or of the frame);
- x_i, y_i are the coordinates of the ith foot of the walking frame, or of the CoP for the ith anatomical foot;
- n is the number of feet in contact with the ground. When all the feet are on the ground, $n = 6$ (2 anatomic feet, 4 frame feet).

Therefore, according to (1), at any instant in time, we must know the magnitude and position of the vertical load acting on each foot of the walking frame and acting on each anatomical foot of the person.

We also define the instantaneous combined BoS to be the convex polygon formed by the boundaries of the anatomical and WF feet in contact with the ground and interconnecting lines between them. Finally, in accordance with the walking robot literature [21], we define the instantaneous stability margin (SM_{inst}) as the shortest distance between the combined CoP and the nearest edge of the combined BoS. It should be noted that, from the definition of CoP alone, it can be proven that, when the CoP reaches an edge of the BoS, the load under all feet, except those forming that edge, will be zero (i.e., when $SM_{inst} = 0$ tipping begins).

Furthermore, we also introduce into our analysis the rate of change of the stability margin. When the instantaneous SM is low, but the rate of change shows that SM_{inst} is rapidly increasing, then it could be concluded that the user is unlikely to fall because they are becoming more stable. Conversely, if the rate of change shows a rapid decrease in the SM_{inst}, then their risk of falling may be higher than SM_{inst} suggests.

Finally, SM_{inst} is likely to be misinterpreted when, for example, SM_{inst} is close to zero because the user is in the process of transferring their body weight from one foot to another that has not yet touched the ground. Conversely, if a foot is in the process of taking off, the user may be less stable than SM_{inst} suggests. Therefore, we also calculate the "projected" stability margin (SMp) which we define to be the shortest distance between the combined CoP and the nearest edge of the "projected" combined BoS. The "projected" combined BoS is calculated post-hoc to be the position of the combined BoS at a point in time t s later. The time t for each individual is the average duration of the terminal swing phase (or landing phase), calculated as 13% of the user's own mean gait cycle duration [27].

2.2. Instrumentation development

To measure the required data, the Salford Walking Aid System (SWAS) was developed consisting of:

(a) A purpose-designed instrumented walking frame (WF) to measure the vertical force acting through each of its legs.

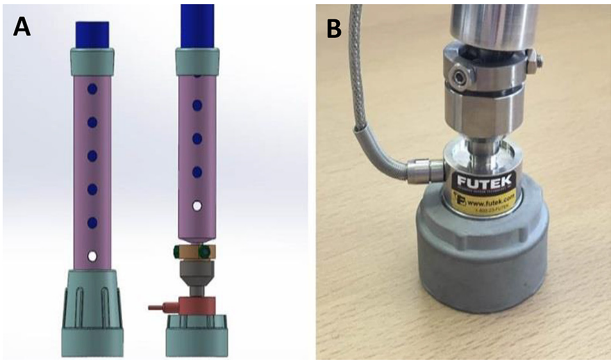

Fig. 1. (A) Model of an instrumented foot of the walking frame with integrated load cell. (B) Instrumented foot with adjustable vertical axis of the load cell configuration, i.e., set to be perpendicular to the ground.

(b) Commercial in-shoe sensors (medilogic®insole, T&T medilogic Medizintechnik GmbH, Schönefeld, Germany) to measure the pressure distribution and hence the resultant vertical force and the corresponding CoP location for each anatomical foot.

(c) An optoelectronic motion capture system to capture the position of both, the anatomical feet and walking frame feet. For this study, a mobile 6 camera system (Qualisys Oqus300, Qualisys AB, Göteborg, Sweden) was used.

The instrumented WF was modified to accommodate a single axis load cell (Futek LCM300, Futek Advanced Sensor Technology Inc., Irvine, CA, USA) in each leg of the frame in order to measure the vertical ground reaction forces (Fig. 1). The force data are sent to a laptop by wireless transmitters (Mantracourt T24-ACMi, Mantracourt Electronics Limited, Exeter, UK) fixed onto the frame. Design requirements included the necessity to be able to adjust the frame height for a range of users, to ensure that the axis of each load cell was perpendicular to the ground during the frame stance phase, and to minimise the weight added to the frame. Currently, the total weight of the instrumentation is 1 kg, which includes load cells, transmitters, batteries, and titanium connectors needed to integrate the load cells into the frame legs. Moreover, to obtain the selected outcome measures, walking frame load cells, pressure insoles, and optoelectronic camera data collection needed to be synchronised; to this end, the Medilogic system was modified to receive a sync pulse to allow for synchronisation of foot pressure data with load cell data and position data.

The instrumented walking frame was first tested using a force plate to verify the accuracy of the load cells: the device was loaded with known weights and the vertical force was measured by each load cell compared to that measured by a force plate. Finally, data from both the SWAS and force plate were recorded from a user picking up the frame, placing it forward onto the force plate, then stepping into it (a large 600 × 900 mm AMTI BP600900 force plate normally used for sprinting was used to allow for simultaneous contact with all 4 WF feet and both anatomical feet). This allowed the COP_{System} calculated from load cell, insole and camera data to be compared against that calculated from force plate data.

2.3. Data processing

In order to process the force and position data, software written in MATLAB was developed to:

- Detect when each of the frame and user's feet are on the ground (supporting feet) through identification of individual touch-down (TD) and lift-off (LO) events of each foot of the frame from load cell data, and TD (i.e., heel-strike) and LO (i.e., toe-off) events of the user's feet from force/insole data; load cell and insole signals were lowpass filtered at 6 Hz with a 4th order Butterworth filter.
- Define the base of support at any time instant as the convex polygon formed by the boundaries of the anatomical and WF feet in contact with the ground and interconnecting lines between them.
- Calculate the resultant vertical force and the corresponding CoP location for each anatomical foot. This uses the individual pressure value from each sensor within each insole, together with the relative position of each sensor in each insole, and the global position of the insole itself.
- Apply (1) using 3D position data, load cell data of the walking frame feet, and the magnitude and coordinates of the resultant load that acts on each anatomical foot (calculated previously).
- Calculate the stability margin as the perpendicular distance from CoP_{system} to the nearest edge of BoS_{system}.
- Calculate the rate of change of the stability margin by differentiating the stability margin curve.
- Calculate the projected stability margin as the perpendicular distance between CoP_{system} and the nearest edge of the projected combined BoS, at a participant-specific instant t s forward in time.
- Calculate device loading as the percentage of body weight transferred by the user onto the device.
- Determine the movement sequence of the frame in relation to the user's foot placements.

2.4. Subjects

One young adult (age = 27) and one older WF user (age = 83) were recruited to test the feasibility of the protocol and to establish proof-of-concept for the method. The older subject met the inclusion criteria of being able to walk household distances with a walking frame but not being able to walk such distances unaided. A description of the participants' basic gait parameters is provided in Supplement A. Written informed consent was obtained and the experimental protocol was approved by the University of Salford Ethics Committee (HSCR13-48).

2.5. Protocol

To test our method, the young adult and the older WF user were asked to walk with the SWAS in a home-setting: the University of Salford Activities of Daily Living (ADL) flat (furnished, and equipped with 6 optoelectronic cameras). Here participants walked 3 times with the SWAS at their self-selected speed from the kitchen to the bathroom (6 m). This pathway was selected as it included two consecutive 90° turns (through two doorways: kitchen to lounge, lounge to bathroom) and transitions between different flooring conditions (vinyl to carpet, carpet to vinyl), therefore representing real-world challenges seen in users' homes. For all trials, subjects were asked to walk with the WF as recommended by clinical guidance: to lift the frame forward and, only once it is grounded, to then step into the frame.

3. Results

3.1. Load cell testing

A Maximum error of 5% and Root Mean Square value of 0.46N were obtained when comparing the vertical force recorded by each load cell to the corresponding data recorded with a force plate. With regard to the accuracy of the CoP_{System} calculated with our sensor system, a maximum error of 25 mm in mediolateral and 17 mm in anteroposterior direction was found, which we consider

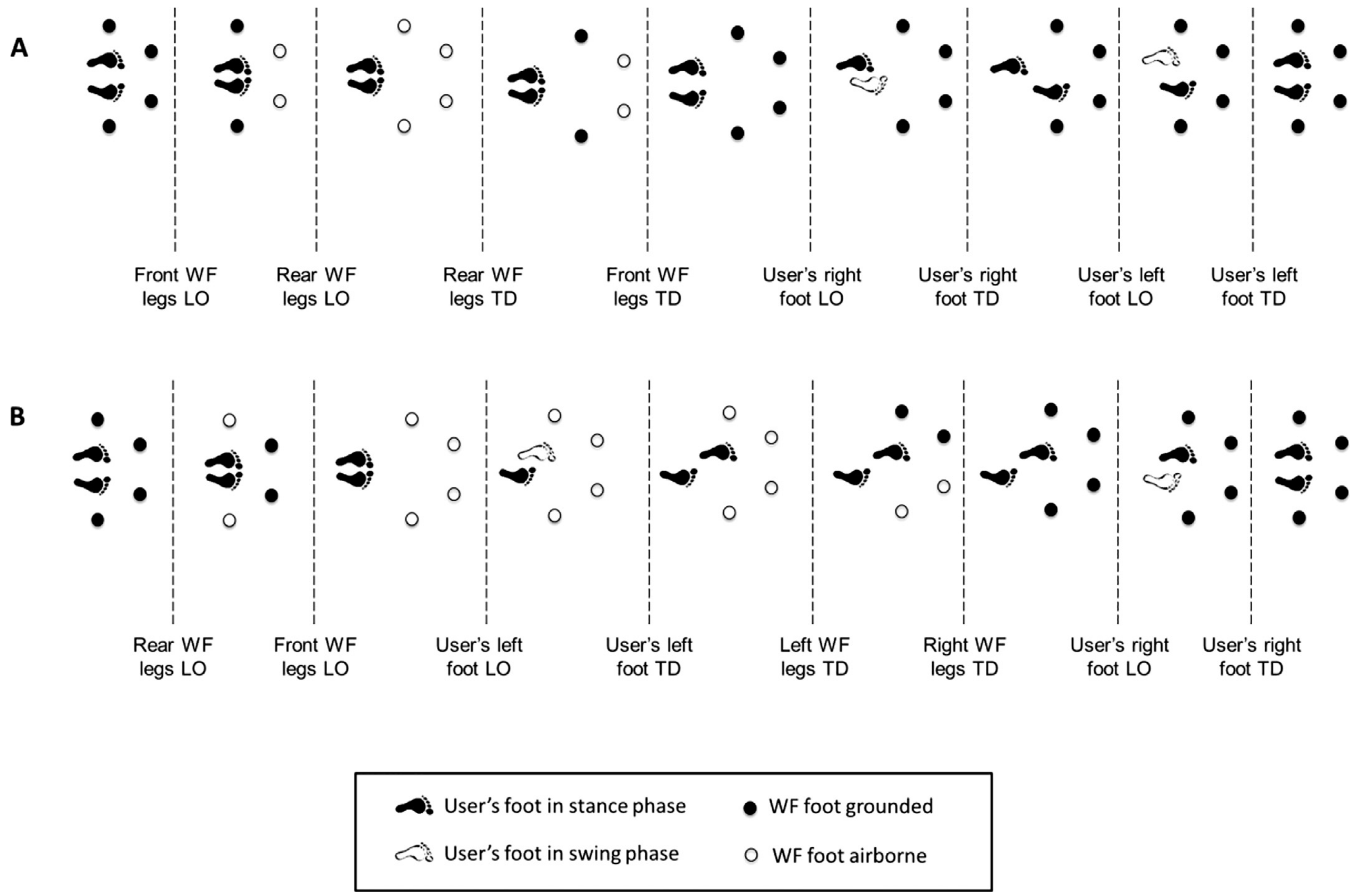

Fig. 2. Examples of lift-off and touch-down sequence of user's and frame's feet for one movement cycle when walking with a walking frame in accordance with clinical guidance (A) and when walking with a phase of single support during which only one anatomical foot is on the ground, followed by a mediolateral rolling of the frame at touch-down (B). Foot prints indicate gait phases (black foot prints indicate stance; white foot prints indicate swing), and dashed lines represent touch-down/lift-off events of anatomical and/or WF feet.

acceptable being equal, respectively, to 4.24% and 2.07% of the maximu*m* width and length of the combined BoS in the testing conditions studied.

3.2. Characterization of foot-ground contact events and movement sequence

Walking with a pick-up walker differs significantly from unassisted walking in that the user needs to coordinate the movements of the device together with their own foot movements. Specifically, TD and LO gait events exist for the feet of both, the user as well as the frame, and the sequence in which they occur may vary greatly from one movement cycle to the next (see Fig. 2).

Figs. 3 and 4 show four examples of the timings of frame movements in relation to foot movements for one gait cycle obtained during straight line walking and one during turning for both the young adult (Fig. 3) and the WF user (Fig. 4). For the purpose of this study, the gait cycle is defined as the period starting when the first WF leg is lifted off the ground and finishing at the following first lift-off event of the WF. According to clinical guidance, Fig. 3 represents an example of correct use as the young user only steps after the WF is firmly on the ground. Similarly, the older WF user demonstrates correct use of the device during straight line walking (Fig. 4(A)), however, during turning (Fig. 4(B)) the older user steps while the WF is still airborne (creating a single support phase). This contradicts clinical guidance and Fig. 8(B) shows that stepping while the frame is airborne greatly decreases the stability margin during that phase.

3.3. Characterization of system stability

Table 1 summarises the number of movement cycles for young adult and WF user in the ADL flat and presents terminal swing phase duration, minimum SM_{inst}, and mean rate of change of SM_{inst} all averaged over the total number of gait cycles for both participants.

Fig. 5(A), (B) and Fig. 6(A), (B) illustrate the CoPs of the frame (CoP_{Frame}), the user (CoP_{User}), and the combined system (CoP_{System}) – each in relation to their respective BoS for two different time instants of the overall movement cycle. Fig. 5(A) shows CoP_{Frame} on the edge of BoS_{Frame}, which, if viewed on its own, would indicate instability. However, this is because the frame is about to be lifted, which is part of the general movement cycle. When looking at CoP_{System} in Fig. 5(B), it becomes clear that it is very close to CoP_{User} and well within BoS_{System} and, therefore, the system is stable even though the frame alone appears unstable. Similarly, in Fig. 6(A), CoP_{User} is near the outer edge of BoS_{User} (right foot single support) due to leaning onto the device, however, as the frame is providing substantial support, CoP_{System} is well within BoS_{System} (Fig. 6(B)) and therefore one can conclude that the overall system is stable, even though the user alone appears unstable.

By graphically representing the variation of the instantaneous stability margin (SM_{inst}) over time, the user's overall stability in

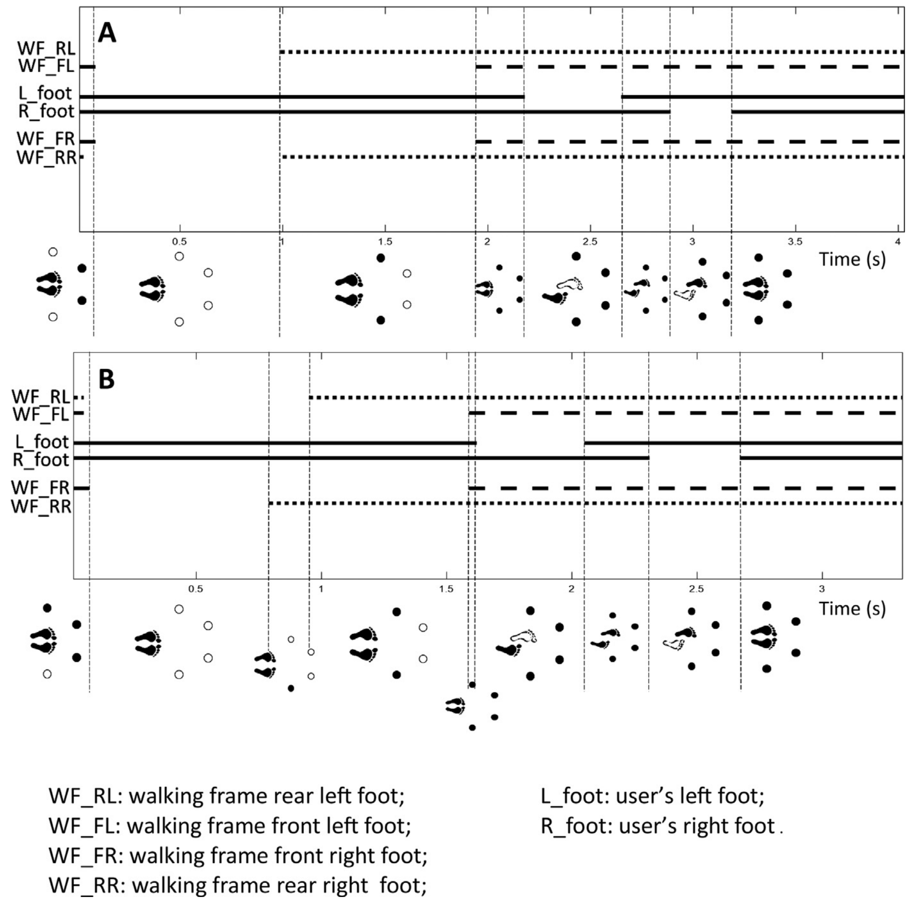

Fig. 3. Movement sequence: horizontal lines indicate times where feet of user and/or frame are grounded. Young adult following clinical guidance to only move themselves once the walking frame is solid on the ground for both, straight line walking (A) and turning (B). Dashed lines represent touch-down/lift-off events of anatomical and/or WF feet.

Table 1
Number of movement cycles and descriptive statistics for the Stability Margin and its rate of change.

	Total number of cycles	Num. straight line walking cycles	Num. mturning cycles	Average cycle duration (s)	Terminal swing phase duration (s)	Min SM_{inst} (mm)	Pos. mean rate of change of SM_{inst} (m/s)	Neg. mean rate of change of SM_{inst} (m/s)
YA	18	8	10	3.42	0.377	77.76	0.24	−0.37
WF user	16	10	6	4.82	0.190	64.9	0.15	−0.23

relation to different movement patterns and walking conditions can be characterized. Fig. 7(A) and (B) respectively illustrate the SM_{inst} for the same straight line walking and turning gait cycles represented in Fig. 3(A) and (B) for the young adult. Similarly, Fig. 8(A) and (B) illustrate the SM_{inst} for the same straight line walking and turning gait cycles represented in Fig. 4(A) and (B) for the WF user. It is evident that, for both the young adult and the WF user, SM_{inst} reaches its maximum when the WF is grounded. During turning however (Fig. 8(B)) the SM_{inst} for the WF user drops to 12.7 mm and approximately 5 times lower than in straight line walking (Fig. 8(A)) and 6.5 times lower than the minimum SM_{inst} for the young adult during turning (Fig. 7(B)). It should also be noticed that, in Fig. 8(B), the SM_{inst} reaches its minimum when the WF user is in single support. Nevertheless, the rate of change of the SM_{inst} indicates that, although the instantaneous SM value during single support appears to be low, it is increasing, suggesting that the CoP_{System} is moving towards a more stable position. Similarly, the projected stability margin predicts that an imminent change of BoS_{System} will cause the SM_{inst} to increase, thereby improving the stability of the system (Fig. 8(C)). In contrast, in Fig. 8(A), SM_{inst} decreases drastically at the onset of the second step reaching the value of 66 mm (45 mm lower than that relative to the same event in Fig. 8(B)), thus suggesting increased instability probably due to the user's posture being excessively upright.

3.4. Characterization of device loading

Fig. 9 shows an increase in device loading for the walking frame user as compared to the young subject during straight line walking: whilst the young adult uses the frame only for light touch support, the older frame user uses it for structural support.

4. Discussion

We have developed and then demonstrated a novel method for the investigation of stability in walking aid users in a standardized home setting. Specifically, we have introduced a novel outcome measure which is generalisable to a range of walking aids, the stability margin of the combined system (user + device), and we have demonstrated that the stability margin of the combined system should be used for making inferences on WF user stability. Our stability margin was adapted from the walking robot literature [21–24] but, in order to take full account of weight, acceleration (linear and angular), and externally applied forces, the CoP

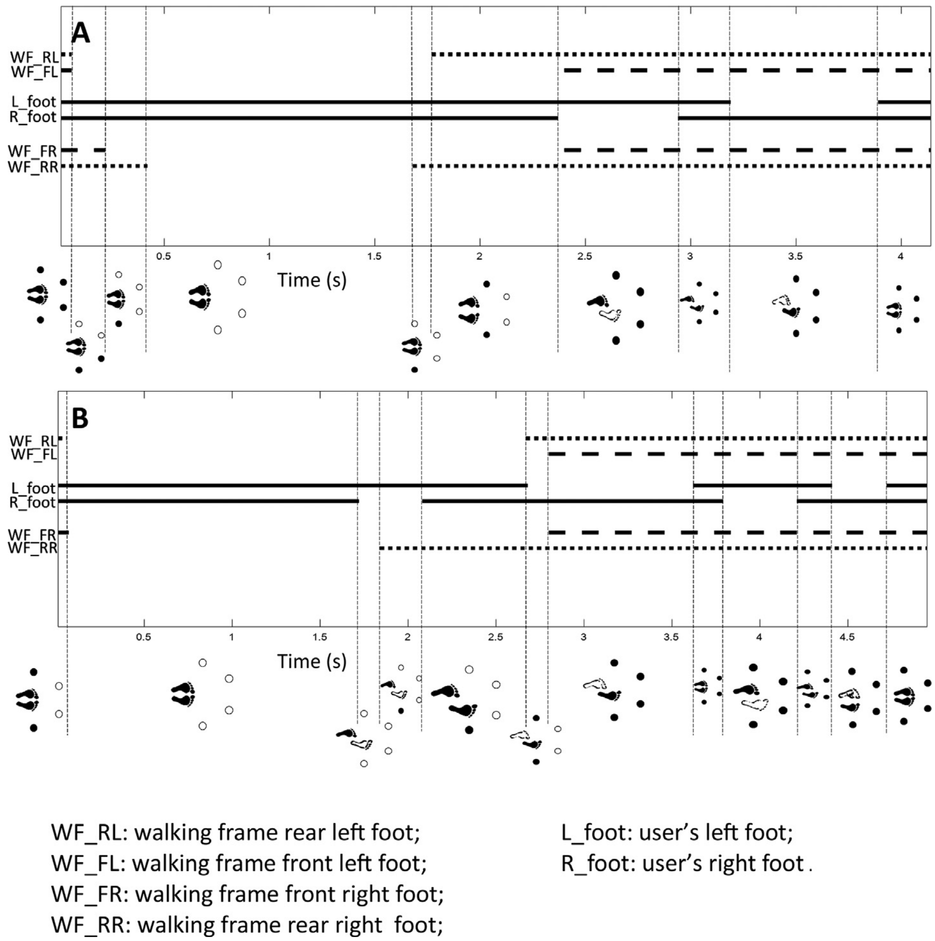

Fig. 4. Movement sequence: lines indicate times where feet of user and/or frame are grounded. Older WF user following clinical guidance to only move themselves once the walking frame is solid on the ground for straight line walking (A) but lifting and moving their right foot whilst the walking frame is still in the air during turning (B). Dashed lines represent touch-down/lift-off events of anatomical and/or WF feet.

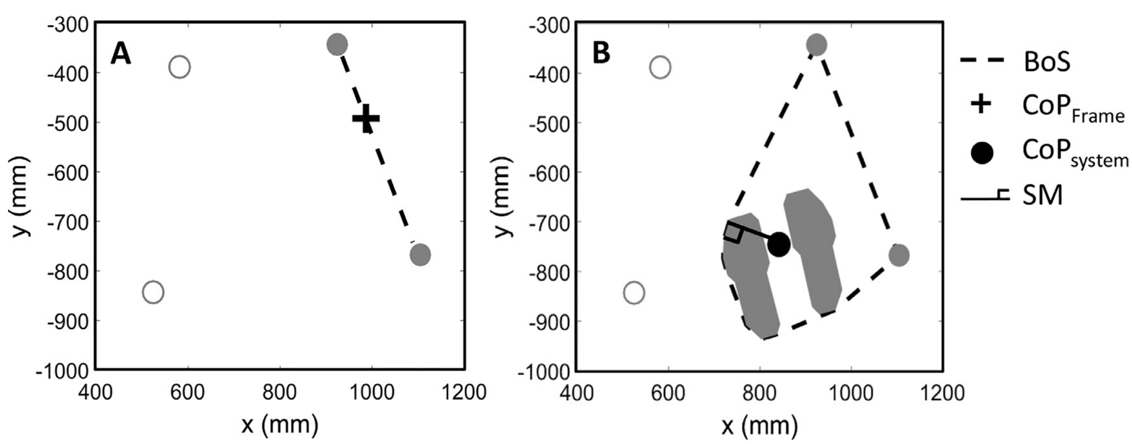

Fig. 5. Illustration of CoP_{Frame} (A) alone in comparison to CoP_{System} (B) for an instant in time during which only two feet of the WF are on the ground, highlighting the importance of CoP_{System} in relation to BoS_{System} for accurate evaluation of the moving system's stability. Grey foot prints indicate stance; white foot prints indicate swing.

was used instead of the vertical projection of the centre of mass. Indeed, although these two measures correspond when there is negligible acceleration and no external forces applied, in dynamic situations, tipping begins when the CoP, not the centre of mass, reaches the boundary of the base of support. Previous authors have investigated the kinematics of the CoP_{WF} and device loading for their inferences on stability and have concluded that, when the WF is on the ground, the user's loading of the device is directly proportional to the risk of falling [19,20]. Their approach did not correctly quantify stability, relied on all four feet being on the ground making it inapplicable to cases when the WF was airborne or in the process of touching-down/lifting-off, and could not distinguish tipping from lifting of the device. Conversely, our approach is able to assess stability during all phases of gait, including when the WF is fully or partially grounded, but also when it is completely airborne (in which case BoS_{System} reverts automatically to BoS_{User}).

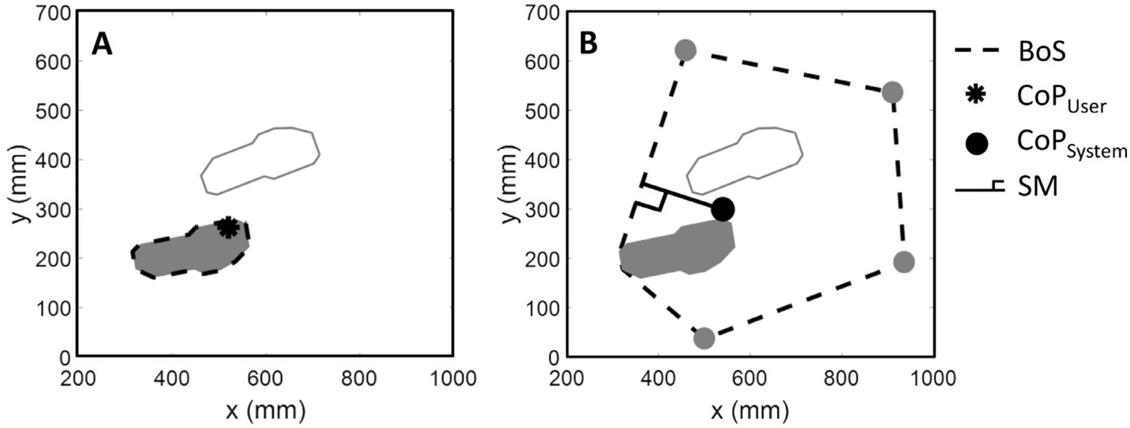

Fig. 6. Illustration of CoP_{User} (A) alone in comparison to CoP_{System} (B) for an instant in time during which only the user's right foot is on the ground and the user is leaning forward onto the WF, highlighting the importance of CoP_{System} in relation to BoS_{System} for accurate evaluation of the moving system's stability. Grey foot prints indicate stance; white foot prints indicate swing.

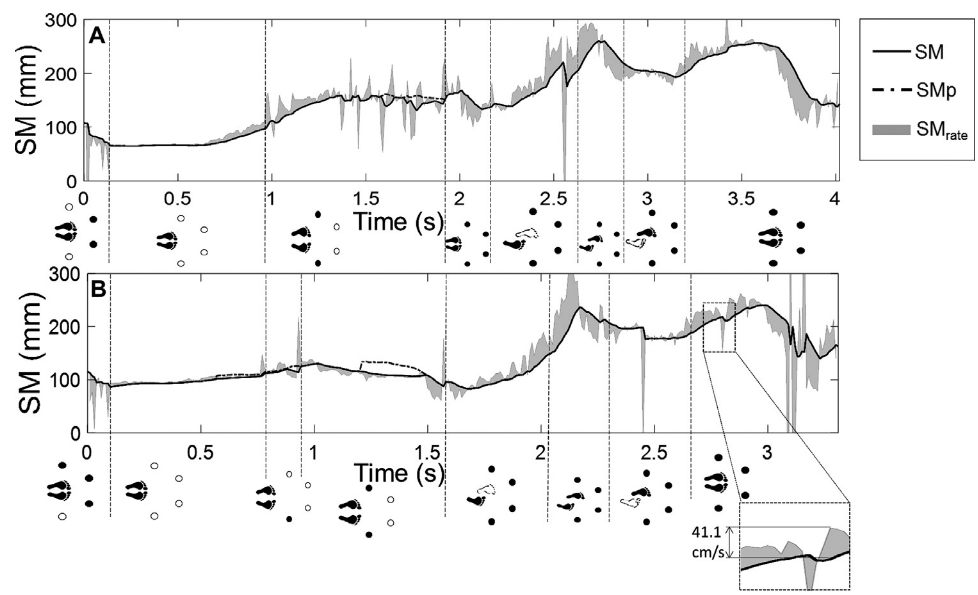

Fig. 7. SM_{inst} over a gait cycle for a young adult (A) walking in a straight line in the ADL flat, and (B) for a 90° turn in the ADL flat. The grey area shows SM_{rate} (i.e., the rate of change of SM_{inst}) in units of cm/s (note that the 0 value has been shifted to lie on top of SM_{inst}), whilst SMp represents the projected Stability Margin, which takes into account not just the feet already in contact with the ground, but also those where touch-down/lift-off is imminent. It can be observed that SM_{rate} presents very high peaks (low troughs) in correspondence to touch-down (lift-off) events of one or more feet: this is due to the instantaneous change in the BoS. For clarity of illustration, only phases that last longer than 0.1 s are represented by footprints.

In this study, to calculate the stability margin, we record load cell, insole, and 3D position data of the frame feet and anatomical feet. Moreover, although the stability margin is normally sufficient to describe WF use, in order to also take into account dynamic situations (e.g., when the user is in single support) or those situations in which a low SM_{inst} does not indicate instability but is due to imminent touch-down of the WF or heel-strike of the next stance foot, the rate of change of the SM_{inst} and the projected SM are also calculated. In addition, our system informs on the user's device loading based on the percentage of their body weight supported by the device. This information is expected to be particularly useful for clinicians who, especially during rehabilitation programmes, recommend their patients to transfer a specific amount of body weight onto the device as this is supposed to optimise the healing and recovery process. However, without any means of measurement, it is extremely difficult for the patient to follow such instructions and for the physiotherapist to evaluate the patient's compliance with these. Finally, the system further informs on the timings of user and device movements individually and in combination with one another to assess whether the movement sequence conforms with clinical guidance. Since our approach is generalisable to other walking aids, including crutches and walking sticks, it opens up significant opportunities to investigate stability in users of other devices.

However, it should be noted that, although the methodology introduced in this paper (i.e., the use of the combined stability margin) is generalisable, device-specific modifications to accommodate the load cells in such a way as to accurately measure vertical ground reaction forces may be required.

Moreover, at this stage, the SWAS is designed to report on stability only and cannot be utilized as a long-term monitoring or fall detection tool. Therefore, although it is able to detect a reduction in stability, it cannot inform on the circumstances that caused such reduction. For this, additional instrumentation such as video

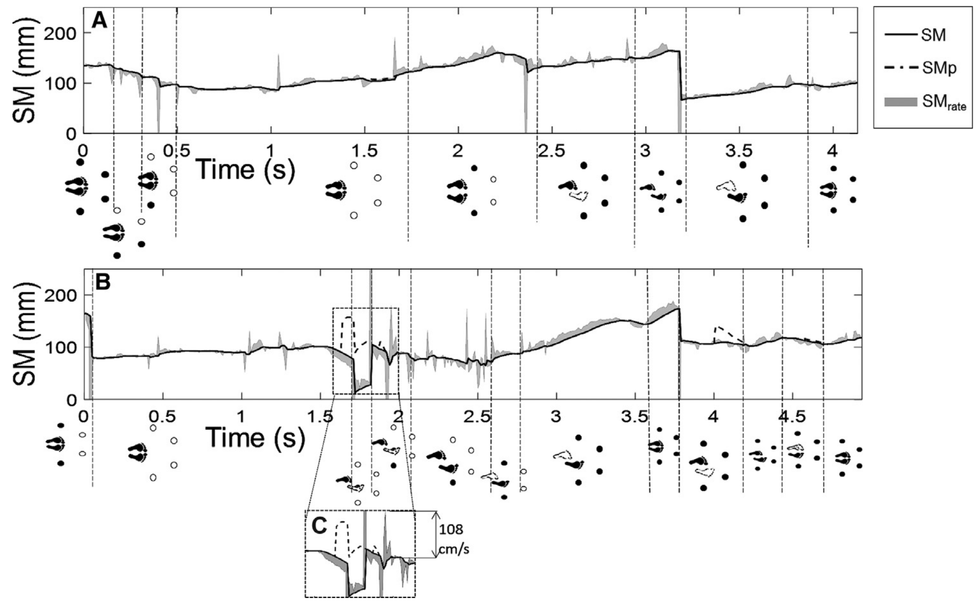

Fig. 8. SM_{inst} over a gait cycle for a WF user (A) walking in a straight line in the ADLflat and (B) performing a 90° turn in the ADL flat. (C) The grey area shows SM_{rate} (i.e., the rate of change of SM_{inst}) in units of cm/s (note that the 0 value has been shifted to lie on top of SM_{inst}), whilst SMp represents the projected Stability Margin, which takes into account not just the feet already in contact with the ground, but also those where touch-down/lift-off is imminent. It can be observed that SM_{rate} presents very high peaks (low troughs) in correspondence to touch-down (lift-off) events of one or more feet: this is due to the instantaneous change in the BoS. For clarity of illustration, only phases that last longer than 0.1 s are represented by footprints.

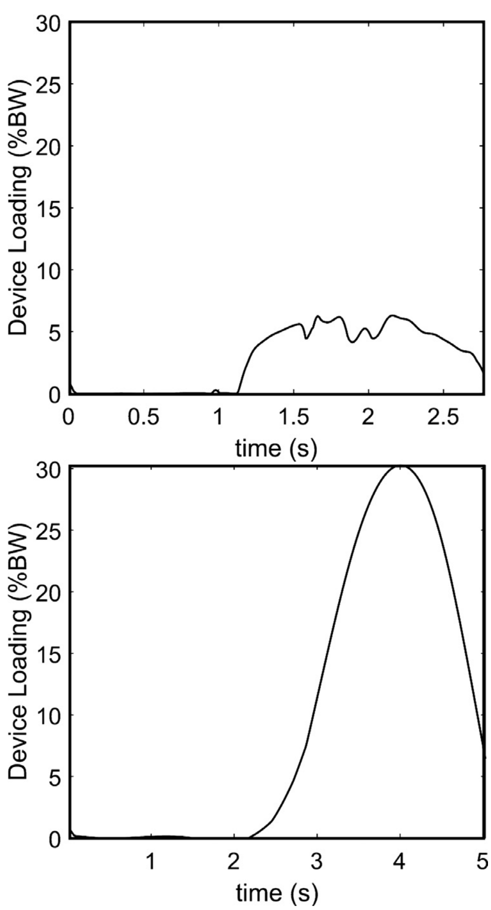

Fig. 9. Device loading: (a) using the frame for light touch support (young adult) versus (b) structural support (WF user) in the ADL flat when the frame is grounded.

cameras or inertial sensors would be needed to identify underlying causes such as collisions of the frame with either the user's feet or objects of the environment.

Finally, to calculate the stability margin, the SWAS system relies on knowing the location of the CoP of each anatomical foot with respect to the frame. At present, we use optoelectronic cameras to obtain the required position data of the anatomical and frame feet, but in future a more portable solution based on dedicated position sensing is required for home use.

Longer term, this method is expected to contribute to improved device prescription, user training and monitoring, and device design, all of which should impact positively on the quality and the frequency of use.

Competing interests

None declared.

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The experimental protocol was approved by the University of Salford Ethics Committee (HSCR13-48).

Supplementary materials

Supplementary material associated with this article can be found, in the online version, at doi:10.1016/j.medengphy.2017.06.013.

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