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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.

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.