Assessment of Balance Instability by Wearable Sensor Systems During Postural Transitions

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Abstract— **Several studies have demonstrated beneficial effects of real-time biofeedback for improving postural control. However, the application for daily activities, which also include postural transitions, is still limited. One crucial aspect is the time point of providing feedback, and thus its reliability. This might depend on the sensor system used, but also on how the threshold is defined.**

This study investigates which wearable sensor system and what kind of threshold is more reliable in a situation of a postural transition. To this end, we compared three sensor systems regarding their accuracy in timing in a stable and unstable Postural Transition in 16 (9 female) healthy young adults: a multiple Inertial Measurement Unit system (*IMU*), a pressure Insole system (*IS*), and a combination of both systems (*COMB*). Further, we contrasted two threshold parameters for each system: a Quiet Standing-based threshold (*QSth*) and a Limits of Stability-based threshold (*LoSth*).

Two-way Repeated Measures ANOVAs and Wilcoxon tests ($\alpha{-}level: 0.05$) indicated highest accuracy in the *COMB LoSth*, though with small differences to the *IS LoSth*. The *LoSth* showed more accurate timing than the *QSth*, especially in medio-lateral direction for *IS* and *COMB*.

Consequently, for providing a reliable timing for a potential biofeedback applied by a wearable device in everyday life situations, such as during postural transitions, application should focus on pressure Insoles and a functional stability threshold, such as the LoS-based threshold.

I. INTRODUCTION

Falls are among the most common causes for injuries in the elderly and their prevalence is further increasing [1]. The main reasons for falls are gait and balance disorders resulting from a physical decline or neurological disease [2]. Consequently, researchers employ sensor technologies to objectively quantify balance disorders [3] and provide sensory feedback helping to improve patients' balance [3][4]. Besides stationary systems wearable devices have been developed. The devices are mostly based on Inertial Measurement Units (IMU) or pressure Insoles (IS) that estimate the patients' body motions [4]. A high variability in the center of mass (CoM) or center of pressure (CoP) trajectory indicate balance instabilities [4]. Once the CoM shifts outside the Base of Support (BoS), the body becomes unstable and there is a risk to tip over [5]. However, humans are able to cope with a certain degree of imbalance in order to prevent falls. A metric to quantify this ability are the so called Limits of Stability (LoS), the maximal displacement an individual can lean in any direction from an upright position without changing the BoS [6]. To measure static balance [7] and to provide biofeedback [4], often a single IMU is attached to the patient's lumbar area as an approximation of the CoM. On the other hand in more dynamic situations, such as during gait, force or pressure sensors are preferred for an accurate gait analysis [7] and for providing biofeedback [4].

For providing biofeedback researchers often either defined a certain inclination angle of the trunk [8], a specific velocity or acceleration of baseline trials [9][10] or a certain CoP displacement [11] as the feedback threshold. While previous works [4] have mainly focused on providing biofeedback for improving postural control during stance and gait, the application for daily activities, which also include postural transitions, such as bending over, is still limited [12]. For the feedback to be successful, it is crucial to define reliable criteria that indicate balance instability, and thus a reliable time point to give feedback. However, the accuracy in timing a potential biofeedback, and thus its reliability might depend on the sensor system used, but also on the threshold definition.

Consequently, the aim of this work is to investigate which wearable sensor system and what kind of threshold is more reliable during postural transitions. Therefore, we compare three sensor systems with each other regarding their accuracy in timing in a stable (sPT) and unstable postural transition (uPT) in 16 (9 female) healthy young adults: a multiple Inertial Measurement Unit system (*IMU*), a pressure Insole system (*IS*), and a combination of both systems (*COMB*). Further, we contrast two threshold parameters for each system: a Quiet Standing-based threshold (*QSth*) and a Limits of Stability-based threshold (*LoSth*).

Commonly a certain angle of the trunk is used as a threshold for providing biofeedback. However this might lead to a feedback in a situation in which the posture itself is stable, such as during postural transitions. Thus, we decided to use a multiple IMU system, which can more accurately estimate CoM [13]. Moreover, in dynamic situations a beneficial effect of biofeedback has only be observed when force/pressure sensors have been used [4]. Moreover, a biomechanical model based on a multiple IMU system can accurately estimate the CoM and ground reaction forces [14], which are commonly assessed by force plates or pressure Insoles [4][7]. Moreover, the choice of threshold might influence the efficacy of a biofeedback given. Thus, e.g. vibrotactile feedback reduced step reaction times in elderly in the work of Asseman et al. [15], however a work by Lee et al. [16] with an equal setup but different threshold

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definition did not show reductions.

II. METHODOLOGY

A. Subjects

16 (9 female) healthy athletic young adults (18-35 years, 1.73±0.1m, 64.5±8.98kg, BMI: 21.49±1.6kg/m^2) without any known medical history of neurological or musculoskeletal diseases participated in this study. Since the available measurement devices were restricted to a body height of 1.50 - 1.95m and a foot size of 23 - 29cm, participation was limited to individuals within these size ranges. The study was approved by the ethics committee of the Technical University of Munich. All participants gave written informed consent.

B. Technical Setup

As the multiple *IMU*-system we use the *Xsens MVN Link* (*Xsens Technologies B.V., Netherlands*) (cf. Fig. 1 (left)), an established system employed in several studies to track the CoM's position [14][17][18].

The *IS*-system (cf. Figure 1, left) consists of two wireless pressure insoles (*T&T medilogic Medizintechnik GmbH, Germany*), based on which we calculate the CoP's position. Each insole employs up to 240 sensors depending on the shoe size. Each sensor unit can measure a pressure between 0.6 to 64 $\mathrm{N/m^2}$ with an accuracy of 5% of the applied force.

Finally, a force plate (*AMTI HPS-SC*) functioned as the reference system (cf. Fig. 1, middle), measuring the CoP and the time point when the subjects alters their BoS (tipping point). All systems capture with a sampling frequency of 100Hz. Data acquisition is done in each system's software. The systems are synchronized by a global trigger signal.

C. Experimental Procedure

The experiment comprised of four parts: two baseline measurements: Quiet Standing (QS) and LoS; and two Postural Transition (PT) conditions: stable (sPT) and unstable (uPT) Postural Transition. For each of the four parts we conducted three trials (cf. Fig. 1, right). Within the LoS measurement and the uPT each trial consisted of one repetition for each direction. Within the sPT each trial consisted of three transitions. The order of the four parts was same for all participants and each started with a 10s *Intro* phase in which participants were asked to stand quietly in an upright position.

The pressure Insoles were calibrated once in the beginning while the participants lifted both feet for 10s. Further, before each trial the force plate was calibrated (zeroed) in an unloaded state, while the IMU-system was calibrated following the software's guidelines with the participant standing on the force plate in the marked position in a N-pose and subsequently in a T-pose. During all measurements participants stood on the force plate in a bipedal stance with the feet hip-width apart. To ensure same foot position across all measurements, it was marked in the beginning of a session (cf. Fig. 1, middle).

First we assessed subjects' body sway during Quiet Standing (QS). We asked the participants to stand as quietly as possible in an upright bipedal stance in the marked position with eyes open and straight gaze for 45s. We then carried out a four-way-leaning test similar to Thomson et al. [19], in which subjects were asked to lean in anterior, left, right and posterior direction separately as far as they could hold the outmost position (LoS) for 3s. In between each direction participants were asked to come back to an upright position for 10s. In the sPT, participants were asked to bend over until 90° for three times per trial with a 10s break between each transition. To standardize the duration of the postural transition, the experimenter provided a rhythm. In the uPT, subjects were asked to first bend over, like in the sPT, then lean in one direction, as in the LoS, however this time until they tip over. This was repeated for each direction within one trial, with a 10s break between each transition.

D. Data Processing

We computed the CoP based on the outputs and geometry of the pressure Insoles [20]. To calculate the body's CoM based on the IMU-system's outputs, we followed the approach described by Hedegaard et al. [17]. The system's software divides the body into 23 segments and outputs each segment's position over time. To calculate the body's CoM position, we reduced these to 16 segments [17]. We first calculated each segment's CoM position based on the proportion parameters and subsequently the body's CoM using the weighted mean of the segments' CoM positions, weighted by their relative mass.

Afterwards we computed the mean and standard deviation (SD) of the detrended CoP's and CoM's trajectory for mediolateral (ML) and anterior-posterior (AP) direction as baseline sway parameters. The *QSth* was then defined by the ellipse area, which was spanned through the points defined in (1) considering the mean (μ) and SD (σ) in each direction averaged across trials of QS. Due to missing measurements in some of the three trials, two trials were used for calculation. The factor $\sqrt{0.0077}$ was derived from Johannson et al. [21], who found this factor to represent the relationship of the baseline sway area and the limits of stability.

$$\begin{aligned} QSthAnterior &= (\mu_y + 2 \cdot \sigma_y)/\sqrt{0.0077} \\ QSthLeft &= (\mu_x - 2 \cdot \sigma_x)/\sqrt{0.0077} \\ QSthRight &= (\mu_x + 2 \cdot \sigma_x)/\sqrt{0.0077} \\ QSthPosterior &= (\mu_y - 2 \cdot \sigma_y)/\sqrt{0.0077} \end{aligned} \tag{1}$$

The *LoSth* was defined by 90% of the four LoS values (AP, ML) [22], averaged across three trials. Consequently, we connected the four points according to the standard ellipse equation (2) and obtained the ellipse area of the QSth and the LoSth for each system (cf. Fig. 2). a and c are the maximum *QSth*- or LoS-values in AP, b and d in ML. In order to check whether a point $P(x/y)$ is inside the ellipse area, the coordinates are inserted into (2). If the result is smaller or equal to one, the point is inside or on the ellipse's border. We applied these thresholds to the data of the sPT's and the uPT's measurements and evaluated whether or at which time point the thresholds were exceeded. In the *COMB* system,

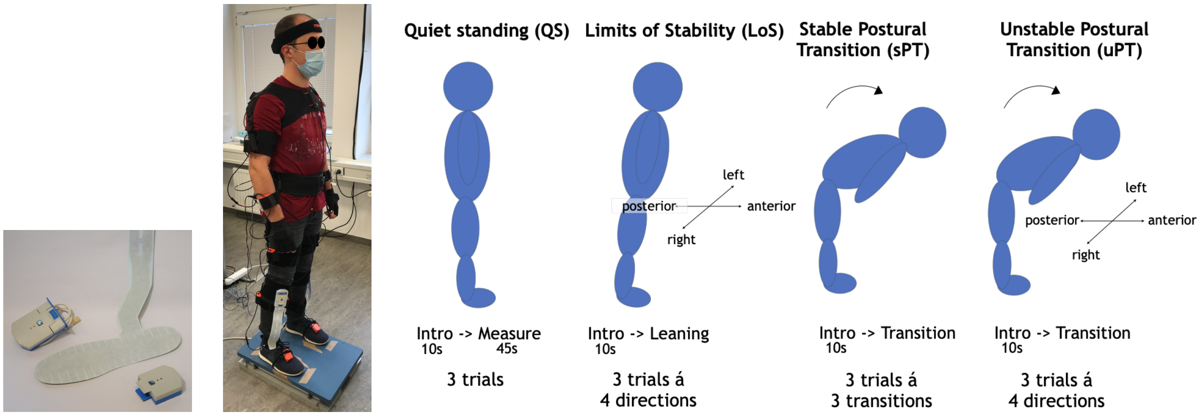

Fig. 1. Left: Medilogic Insoles; Middle: Experimental setup - Participant standing in the initial upright position on a force plate with marked foot position, IMUs (orange boxes) attached via Velcro straps, and pressure Insoles with wireless transmitter in the shoes; Right: Overview of the experimental procedure with two baseline measurements (QS, LoS) and two postural transition conditions (sPT, uPT).

the thresholds were exceeded, as soon as the thresholds of both systems, the *IMU* and the *IS*, were exceeded.

$$\frac{(x-(b_x-d_x))^2}{(|b_x|+|d_x|)^2}+\frac{(y-(a_y-c_y))^2}{(|a_y|+|c_y|)^2}=1 \tag{2}$$

For the sPT we computed the number of thresholds exceeded averaged across total number of transitions multiplied by 100 (proportional threshold exceeded (%)). For the uPT, we evaluated the time discrepancy (Δt) between the tipping point and the time point each system's threshold was exceeded. The tipping point was defined as the time point 20ms before the vertical force (F_z) falls below 85% of the subject's body weight (cf. Fig. 3). shorter Δt indicate a more accurate and thus more reliable timing. The average of the three trials of each subject was used for statistical analysis.

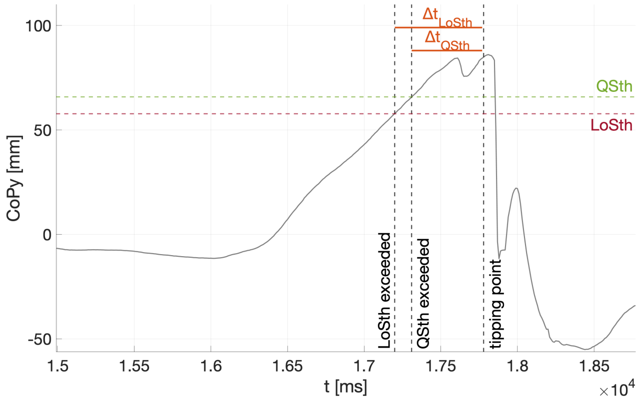

Fig. 3. Visualization of the time discrepancy (Δt) between the tipping point and the point the respective threshold was exceeded

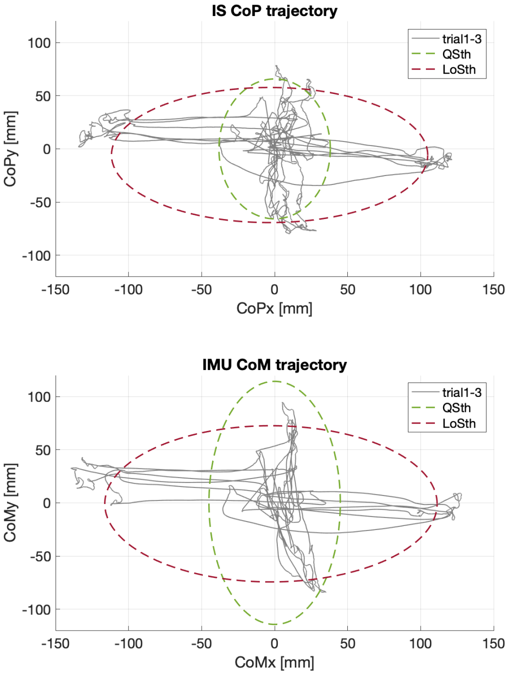

Fig. 2. CoM/CoP trajectories during the LoS measurements and the respective *QSth* and *LoSth* of each system.

E. Statistical Analysis

To assess differences between systems and thresholds, we computed a two-way repeated measures ANOVA following a 2x3 within subjects design with the factors 'Threshold' and 'System' and post-hoc paired t-tests for log-transformed Δt for uPT. Due to not normally distributed data of proportional threshold exceeded and violation of homoscedasticity in the sPT, as well as small sample size ($n = 16$), we computed Friedman and Wilcoxon tests in uPT. Dunn-Sidak correction was applied for post-hoc comparisons. Data processing and statistical analysis were performed in *Matlab (v2020a)*.

III. RESULTS

A. Stable Postural Transition

The two-way Friedman test resulted only in a statistical tendency ($p = 0.059$). However, due to small sample size, we further computed post-hoc comparisons for the factor 'System' and the factor 'Threshold'. One-way Friedman tests revealed a significant difference between systems only in the *LoSth* ($Chi^2(2) = 11.46, p = 0.003, n = 16$).

Pairwise comparisons of systems (cf. Fig. 4) resulted in a significantly lower proportional threshold exceeded in the *IS LoSth* ($M = 0.22, SD = 0.33, p_{corr} = 0.037$) and *COMB LoSth* ($M = 0.22, SD = 0.34, p_{corr} = 0.0147$) than in the *IMU LoSth* ($M = 0.67, SD = 0.39$). Pairwise comparisons of thresholds resulted in a tendency for a lower proportional threshold exceeded in the *IS LoSth* compared to the *IS QSth*.

B. *Unstable Postural Transition*

For the uPT, we evaluated the repeated measures ANOVA separately for each direction (cf. Fig. 5 & 6). Because all values lie above zero, the participants crossed all thresholds before reaching the tipping point. In anterior direction (cf. Fig. 5, left), we obtained a significant main effect of the factor 'System' ($F(2, 30) = 9.4, p = 0.0025, \eta_p^2 = 0.39, f = 0.79$) and a significant interaction effect 'System x Threshold' ($F(2, 30) = 18.66, p = 0.0000, \eta_p^2 = 0.55, f = 1.12$). Post-hoc pairwise comparisons resulted in a significantly shorter Δt for *IS* and *COMB* compared to *IMU* in the *LoSth*, as well as in a significantly shorter Δt for *COMB* compared to *IS* and *IMU* in the *QSth* (cf. Fig. 5, left).

In posterior direction (cf. Fig. 5, right), we found a significant main effect of the factor 'System' ($F(2, 30) = 11.49, p = 0.0025, \eta_p^2 = 0.43, f = 0.88$), as well as a tendency for a main effect of the factor 'Threshold' ($F(1, 30) = 4.43, p = 0.0526, \eta_p^2 = 0.23, f = 0.54$) and interaction 'System x Threshold' ($F(2, 30) = 3.42, p = 0.0649, \eta_p^2 = 0.19, f = 0.48$). Post-hoc tests revealed a significant difference of 1530ms between *IS* and *COMB* in the *QSth* and of 654ms in the *LoSth*, as well as a significantly shorter Δt for *IMU* compared to *IS* for the *QSth*. Regarding threshold effect, we observed a shorter Δt for *LoSth* for *IS*.

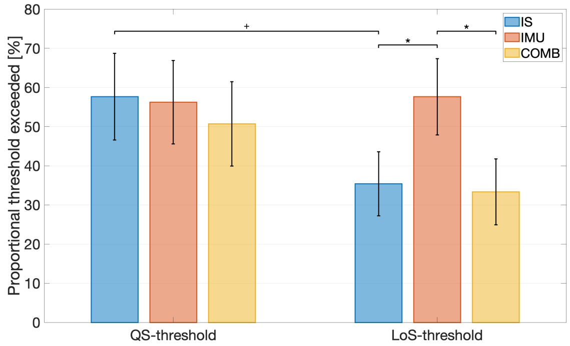

Fig. 4. Proportional threshold exceeded averaged across subjects; whiskers indicate Standard Error (SE), + = p<0.1, * = p<0.05, ** = p<0.01.

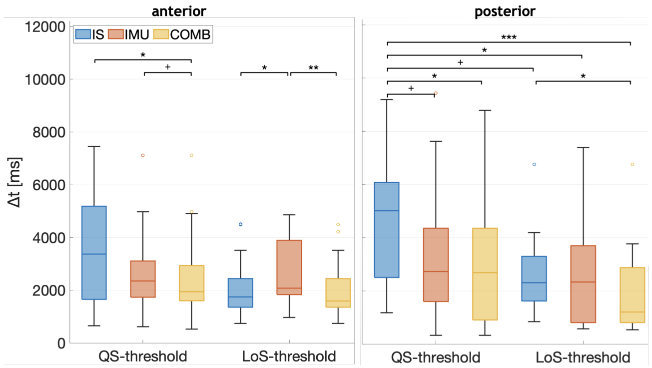

Fig. 5. Time discrepancy Δt between the tipping point and the time point of threshold exceeded during the uPT anterior & posterior across subjects; red line = median, whiskers = min/max, + = p<0.1, * = p<0.05, ** = p<0.01, *** = p<0.001.

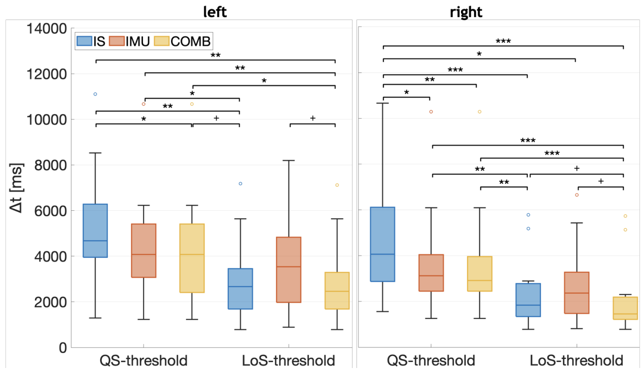

Fig. 6. Time discrepancy Δt between the tipping point and the time point of threshold exceeded during the uPT left & right across subjects; red line = median, whiskers = min/max, + = p<0.1, p<0.05, ** = p<0.01, *** = p<0.001.

During the uPT left (cf. Fig. 6, left), differences between the systems and thresholds were more distinct. Here, we found a significant main effect of both factors 'System' ($F(2, 30) = 5.32, p = 0.0235, \eta_p^2 = 0.26, f = 0.6$) and 'Threshold' ($F(1, 15) = 17.00, p = 0.0009, \eta_p^2 = 0.53, f = 1.06$), but also a significant interaction effect 'System x Threshold' ($F(2, 30) = 11.35, p = 0.0019, \eta_p^2 = 0.43, f = 0.87$). Post-hoc comparisons revealed in the *QSth* a significant difference between *IS* and *COMB*, with shorter Δt for *COMB* and a tendency for a difference between *IMU* and *COMB* in the *LoSth*. Further, a threshold's effect was significant for the *IS COMB*.

Results of the uPT right (cf. Fig. 6, right) were similar: There was a significant main effect of both factors 'System' ($F(2, 30) = 6.56, p = 0.0159, \eta_p^2 = 0.3, f = 0.66$) and 'Threshold' ($F(1, 15) = 39.25, p = 0.0000, \eta_p^2 = 0.72, f = 1.62$), and the interaction effect ($F(2, 30) = 10.14, p = 0.0016, \eta_p^2 = 0.4, f = 0.82$). Within the *QSth*, the *IMU*'s and *COMB*'s Δt was significantly shorter than the *IS*'s, and within the *LoSth* there was a tendency for a shorter Δt for *COMB* compared to *IS* and *IMU*. Finally, regarding a threshold's effect, Δt of the *IS* and *COMB* was significantly shorter in the *LoSth* than in the *QSth* by 2326ms and 1613ms, respectively.

IV. DISCUSSION

This study aimed to investigate which wearable sensor system and what kind of threshold is more reliable during postural transitions. In the next sections we first discuss the influence of sensor system and subsequently the influence of threshold parameters on the accuracy in timing.

A. *Influence of Sensor Systems*

In the *sPT* differences between systems were dependent on the threshold parameters, since we could only observe differences for the *LoSth*. Thereby, *IS* and *COMB* showed highest reliability, due to a lower number of thresholds exceeded (cf. Fig. 4).

In the *uPT COMB* consistently had shorter time discrepancies than the other systems, when the *LoSth* threshold was used. However, absolute differences were small between *COMB* and *IS*, but larger between *COMB* and *IMU*. Differences between *IS* and *IMU* were only significant within the *LoSth* in anterior direction. However, the trend shows that within the *LoSth* time discrepancy for *IS* was shorter and variation between subjects smaller (cf. Fig. 5 & 6).

Average reaction times on vibrotactile stimuli lie around 500-600ms in a dual task setting and around 200-450ms without multi-tasking [23], depending on the age. In our best case, the *COMB LoS*, the average time discrepancy was 1768ms (average across subjects and directions). This is four to five times higher compared to reported reaction times to both a surface perturbation [15][16] and to a vibrotactile stimulus [23]. Consequently, it might still trigger a potential biofeedback too early. However, in extreme cases, time discrepancy between threshold exceeded and the tipping point fell below 600ms, which might be too late for being able to respond in time, if being involved in another task. This high inter-individual variability of our results and the reported interpersonal variability in reaction times [24] and perceptual sensitivity [25], especially with increasing age, suggest that this needs to be considered when designing wearable devices for improving postural control in daily life.

As described in the introduction, we used a multiple IMU-system because previous studies showed that it more accurately estimates the body's CoM [13]. However, the results of this work indicate that the plantar pressure system (Insoles) was still more reliable.

B. *Influence of Threshold Parameters*

Considering differences between threshold parameters in the *sPT*, the *LoSth* tended to be more reliable for *IS* compared to the *QSth* (cf. Fig. 4). In the *uPT* differences between systems and thresholds were more apparent in the ML direction (cf. Fig. 6). The *LoSth* was more accurate than the *QSth*, especially for *IS* and *COMB*. Considering Figure 2, when discussing about threshold differences, it can be observed that the *QSth* showed a more narrow threshold in ML direction compared to the *LoSth*, which was the case for most of the subjects. This can be attributed to a lower baseline sway in ML directions than in AP, due to the chosen hip-wide bipedal stance and because the *QSth* is directly proportional to the baseline sway. However, previous studies have found that body sway increases with age [26][27], while the stability boundary (LoS) decreases [19][21][28]. Consequently, a tighter threshold would be needed with increasing sway for an elderly population. However, besides age, also other factors, such as anthropometry [29][30][31] influence our body sway. Respectively, it has been shown that body sway increases with body height and weight [29][30][31] which can be explained by the inverted pendulum model [32]. This increased body sway would not necessarily be related to an increased risk of falling, since functional limits of stability is also increased with anthropometry [33]. In case the age effect would be neglected, due to a study population of healthy young adults, as in our study, a wider *QSth*, thus, is appropriate when considering the influence of anthropometry.

On the other hand, the LoS has been reported to better consider the individuals' voluntary ability to control balance [6], and therefore might be more relevant in voluntary movements, such as bending over. Moreover, Johansson et al. [21] and Kilby et al. [28] pointed out the importance of the individual's LoS, when determining balance and risk of falling, and goes in line with the previously mentioned relationship of the range of motion and age [28]. However, assessing the real LoS in elderly might be limited, e.g. due to increased fear of falling, and thus might not represent the "real" stability limits [21][34]. Consequently, "comfortable" LoS measurements, which approach the real LoS, might be an alternative for the elderly. Instead of considering 90% of the LoS, as done in this study, e.g. 110% of the "comfortable" LoS could be used.

Limitations and Future Research

As pointed out in previous studies [19][35], the exact measurement of baseline sway and LoS would need more trials to account for possible learning effects. Since our work did focus on the comparison of the systems and threshold parameters and due to the already quite long test duration of 1.5-2 hours, we conducted only three trials of each measurement. Due to the higher inter-individual variability in the *QSth*, future studies consequently should (1) include more than two trials in their analysis, which was the number of trials in our work (in QS) due to missing values in some of the three trials; (2) use eventually less individual-specific parameters (e.g. spectral parameters) [36], or (3) normalize a baseline threshold [31][33] by known factors influencing body sway, such as age [26][27] and anthropometry [29][30].

Moreover, the proportion factor between body sway during QS and the LoS (cf. Eq 1) might be not completely representative for our cohort because the subjects in [21] were older adults and showed smaller LoS than the ones we measured. Thus, future works should further investigate the relationship between quiet standing and the Limits of Stability in different populations, such as in various disorders with disturbed postural stability and different age groups.

Finally, the optimal time span a patient needs to react on the feedback signal should be further investigated under various daily situations and conditions.

V. CONCLUSIONS

The time point when feedback is given is crucial for a potential biofeedback to be successful and reliable. This work investigated how different wearable sensor systems and thresholds affect the accuracy in timing/triggering a potential biofeedback during a stable and unstable Postural Transition.

A combination of a multiple IMU system and a pressure Insole system using a Limits of Stability-based threshold was most accurate in detecting postural instability. However, differences between the combined and the IS system were small. The Quiet Standing-based threshold's definition in our study was not optimal and did not fully reflect the participant's ability to voluntarily control balance in mediolateral direction. Consequently, alternative parameters (e.g. spectral or normalized sway parameters) might provide a better performance. Our results as well as other studies showed that the voluntary ability to control balance is important in the context of postural instability and risk of falling.

Future applications should focus on mobile plantar pressure systems, because the pressure Insoles are less obstrusive and easier to integrate in everyday life than a combination with a multiple IMU system. Moreover, future works should approach a functional stability-based threshold, such as the LoS-based threshold. In the end a prospective device would need to be tested in a cohort of older adults and different daily situations, other than the bipedal stance, to develop a reliable biofeedback system.

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