Understanding Jump Landing as an Oscillating System: A Model-based Approach of Balance and Strength Analyses

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Abstract: Counter movement jumps (CMJ) are well-suited to measure the muscle power and balance. Since it has been clarified that well accepted CMJ amplification-based balance measures (such as TTS or CoP) are significantly influenced by algorithmic, and measurement settings and thus, measurement results have limited meaningfulness among force platforms, we introduce a new model-based approach measuring the postural stability. In this, during the landing and recovery phases after vertical jumps, the lower extremities can be represented by an oscillating system and the corresponding transfer function is described by a second-order delay (PT2) element.

In an initial prospective study with 20 subjects aged over 70 years, we observed an inverse relationship between the calculated parameter w and the jump height and could also identify an influence of sex, and body weight on the jump height. Furthermore, we also found a relation between the parameter w and the dynamic postural stability index (DPSI), even though these results must be ensured statistically using a larger cohort, due to the current limited number of subjects.

Nevertheless, we could confirm the general applicability of the Systems and Control Technology perspective on describing human movements in a potentially more robust manner than current amplification based approaches. Further investigations on our model and the oscillating behavior in the phase of landing are needed to improve our system and to interpret the calculated parameters in a technical and physiological point of view.

1 INTRODUCTION

Geriatric assessments are well-established instruments to identify early changes associated with functional and cognitive decline, as they can occur in common geriatric syndromes, such as frailty or sarcopenia (Clegg et al., 2013; Cruz-Jentoft et al., 2010; Elsawy and Higgins, 2011). Thus, the assessments gain increasing relevance with the ongoing age-related demographic shift. Therefore, it exists a strong research interest to identify degrading abilities very early in geriatric assessments or with technical monitoring systems (Hein et al., 2010; Fudickar et al., 2012), like for example with systems in domestic environments to identify changes in the user behavior (Steen et al., 2013), to trigger preventive measures.

Nevertheless, for a self-determined healthy life and low fall risk, functional abilities and physical fitness are fundamental for healthy aging. Muscular strength of the lower extremities, balance, and endurance are essential factors (Granacher et al., 2013) for the fall risk, frailty, and sarcopenia.

Due to the relevance of muscular strength of the lower extremities, postural stability, and endurance, these factors are covered by various standardized assessments and tests (see Table 1). Most of these assessments and tests consist of several assessment items. For example, the Short Physical Performance Battery (SPPB) consists of a walk test, a static balance test, and the chair rising test and can cover strength and balance only in the combination of the assessment items. Consequently, among the common assessments, only the Counter Movement Jump (CMJ) is well-suited to test both components, strength, and balance within a single item. In detail, the CMJ allows to measure postural stability (balance) (Granacher et al.,

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Table 1: Selection of assessment items in our geriatric study (Hellmers et al., 2016), their test duration, and classification regarding the components of physical fitness (- none, + to +++ increasing significance). The test durations are based on literature and estimated on own experiences (*) in a study with 250 participants. The values in brackets are the durations with introduction and instructions or a test jump.

Geriatric Assessments	Balance	Strength	Endurance	Test Duration
de Morton Mobility Index	++	-	-	9 min
-static balance	++	-	-	
-dynamic balance	++	-	-	
Short Physical Performance Battery	++	++	-	15 min
-static balance	++	-	-	
-chair rise test	-	++	-	
Frailty Index	-	++	-	10-17 min*
-grip strength	-	++	-	3-5 min
Stair Climb Power Test	-	++	-	2-(5) min*
6 Minute Walk Test	-	+	++	6 min
Counter Movement Jump	++	++	-	5-(6) min*

2013) via the time to stability (TTS) during the landing and stabilization phase and muscle performance of the lower extremities (strength) via muscle power ahead of jumps (Buehring et al., 2015; Rittweger et al., 2004; Dietzel et al., 2015; Kalyani et al., 2014).

The assessment of patients' functional status for balance and muscle strength through a single test item instead of several tests lowers costs and personal efforts (supporting an increased assessment density) and can reduce stress and potential fatigue for patients, which hold the risk that assessments results lose significance (Siglinsky et al., 2015). Consequently, the CMJ is a well-suited assessment item to cover both muscle strength of the lower extremities and postural stability.

However, since relying just on the force amplification, current CMJ-based balance measures (such as COP and TTS) have been shown to be significantly prone to algorithmic and technical variations thus have limited viability regarding the generalization of classification and measurements.

Thus, we propose a robust approach to measure postural stability based on the natural frequencies during the landing and recovery phase of CMJs and evaluate its practicability for 20 subjects with an age of 71 to 82 years.

2 COUNTER MOVEMENT JUMP

Within this section, the characteristics of CMJs are discussed and are followed by a description of the technically supported measurement of ground reaction forces.

2.1 Biomechanical Characteristics

Counter movement jumps (CMJs) are vertical jumps that are performed from standing, and according to (Palma et al., 2008) consist of the following phases (as shown in Figure 1): In the first phase (a) the participant is standing. Phase (b) is characterized by the preparation (b) with a downward movement by the flexion of the knees and hips, followed by an immediate and impulsive extension of the knees and hips again to jump vertically up and take-off (c) and flight (d). At the end of the jump, a stage of landing (e) with the absorption of the forces of the impact, and a stage of recovery (f) of the balance can be identified, followed by a standing phase after compensation of the forces (a).

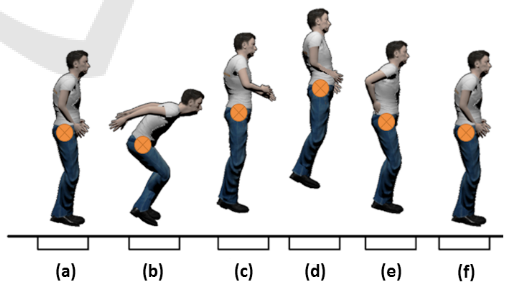

Figure 1: Counter movement jump and its separate phases: standing (a), preparation (b), take-off (c), flight (d), landing (e), and recovery (f). The marks indicate the participant's center of mass in each jump phase.

2.2 Common Technical Measurements via Force Platforms

Since pure observational evaluations are difficult, due to the fast progress of a jump, analyses are typically technology supported via force platforms, contact mats or optical systems (Bui et al., 2015). Force platforms measure the ground reaction force intensity and distribution.

Due to the specific force distribution characteristics, the phases of a CMJ can be identified by variations of the ground force (if measured by ground force reaction platforms or similar devices), as shown exemplarily in Figure 2 and discussed in the following: The transition from standing or rest (a) to preparation (b) can be clearly recognized by changes in the amplitudes of the force, which was nearly constant during the standing phase. Take-off (c) is characterized by the decrease of force affecting on the platform. During the flight phase (d), the force amounts to zero. In the moment of landing (e), the force increases to a maximum. In the phase of recovery (f), the subject tries to compensate the forces, in order to enter the phase of standing or resting (a) again.

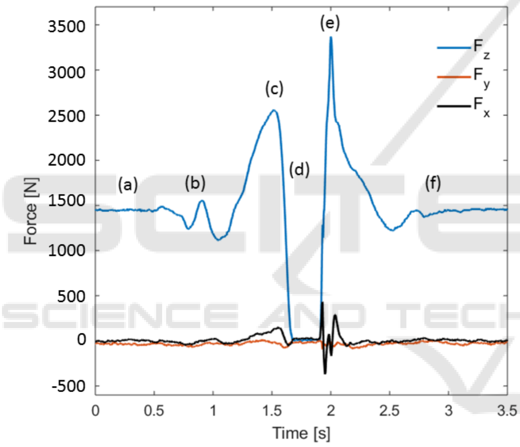

Figure 2: Variations over time of force intensities per axis during a counter movement jump. The force was measured by a force platform. The phases of the counter movement jump are marked in the graph: standing (a), preparation (b), take-off (c), flight (d), landing (e), and recovery (f).

Figure 3 shows the coordinate orientation of the force platform. It is clear that the mean force acts perpendicular to the force plate in the vertical (z) direction. But there are also reaction forces in the mediolateral (y) and especially in the anteroposterior (x) direction. In the anteroposterior direction is a peak during the take-off phase pushing the feet off the ground, and during the landing while the toes and heels strike the ground and compensate the movements.

Force platforms can estimate power, velocity and the related jump height (Samozino et al., 2008). The physical relationships between the measured force and the power, as well as the jump height, is shown in Section 3. The peak force measurements during the preparation and take-off phases e.g. of CMJs are, as long as being considered relative to the body mass, significantly related to muscle strength (Nuzzo et al., 2008; Markovic et al., 2014). The strength can be analyzed by the take-off phase of the jump, and the balance was shown to be estimated based on the force-intensities and distributions during the landing and recovery phase of jumps. For example, the time to stabilization (TTS) is the time it takes for an individual to return to a stable state following a jump or hop landing, and it is a used factor for balance analyses. Thereby, a longer TTS indicates more difficulty controlling the posture of landing and might indicate impaired neuromuscular control (Fransz et al., 2015).

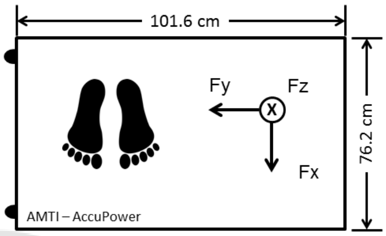

Figure 3: The dimensions of the used AccuPower ground reaction force platform of AMTI with its coordinate orientation. The footprint symbolizes the orientation of the subjects during the jumps.

3 STATE OF THE ART

The jump power, as an indicator of muscle strength, can be identified by the force measurements during the vertical jump, especially in the phase of the take-off.

Force plates measure the force acting on the plate. According to Newton's second law, the force F is equal to the mass m of an object times its acceleration a.

$$F = m \cdot a \tag{1}$$

In the example of Figure 2, we can clearly see the influence of the mass of the jumper on the force F_z acting perpendicular to the surface of the plate. The offset at rest amounts about $1500N$, which corresponds approximately to a mass of $150kg$.

The power P is defined by the force F times the velocity v:

$$P = F \cdot v \tag{2}$$

In many studies, the maximum jump power is observed and seems to be a sensitive indicator of the muscle performance and the strength (Dietzel et al., 2015; Kalyani et al., 2014).

A further important parameter for jump analyses is the jump height. The jump height h can be estimated by the following equation:

$$h = (v_t \cdot t) - \frac{1}{2} \cdot g \cdot t^2, \tag{3}$$

where v_t is the vertical velocity of the center of mass of the jumper at take-off, t is the time to peak flight and g the gravity.

Besides measuring muscle strength, force platforms can be utilized to measure dynamic postural stability, which has been shown as related to balance and ankle stabilities. Therefore, functional deficits such as chronic functional ankle instability (FAI) (Hertel, 2002), can be indicated based on the recorded vertical, anteroposterior or mediolateral reaction forces, which enable the calculation of time to stabilization (TTS) and variations over time of the center of pressure (COP), range of motion (ROM), and the dynamic postural stability index (DPSI) as accepted measures for postural stability and FAI. The DPSI is at least as accurate and precise as TTS but provides a comprehensive measurement of dynamic postural stability that is sensitive to change in 3 directions.

DPSI provides a comprehensive measurement of dynamic postural stability that is sensitive to change in all directions since combining three (vertical, anteroposterior and mediolateral) stability indexes and considers as well the subject's weight for the vertical stability and thus has been shown to be a reliable measure (Wikstrom et al., 2005b; Meardon et al., 2016). While COP and ROM have shown mixed correlations to FAI stabilities, TTS is a well-accepted measure to quantify performance. Typically, the force is considered in order to measure the TTS, as a measure of the ability to stabilize posture (which is applied within numerous studies). TTS typically ranges from 0 to 7s. By investigating 20 TTS calculation methods (as identified via a structured literature review), Fransz et al. (Fransz et al., 2015) have shown that all use threshold-based approaches based on the ground force and 90% can be described based on four aspects: (1) the input signal, (2) signal processing, (3) the stable state (threshold), and (4) the definition of when the (processed) signal is considered stable.

Wikstrom et al. identified a significant variability among TTS measurements due to differences between the TTS calculation methods used in various studies (Wikstrom et al., 2005a). By evaluating the influences of parameter variations, Fransz et al. (Fransz et al., 2015) have indicated that the TTS measure does produce non-standardized parameters if estimated via ground forces reaction parameter. They indicated variations of the TTS of up to 56% for sample rate (100 to 1000 Hz), 37% for filter settings (no filter, 40, 15 or 10 Hz), 28-282% for trial lengths (20, 14, 10, 7, 5 and 3 s), as well as calculation methods. Thereby they clarified the difficulties to compare TTS results recorded among different systems based on the power measure.

While these analyses are performed based on single jump measurements for 25 healthy younger adults (20-53 years), its insights will generally apply due to the indicated computational differences and the drastic effect sizes.

Consequently, alternative measures are desired, which are more robust regarding measurement-variations such as sample rates.

Ideally, these measures should be equally applicable to rather mobile measurement devices such as inertial measurement units (IMU), which will be increasingly applied due to their lower price and the higher grade of mobility (Choukou et al., 2014; Elvin et al., 2007; Milosevic and Farella, 2015).

4 SYSTEMS AND CONTROL TECHNOLOGY

Considering the situation, that a system is stimulated by an action (input signal). Usually, the system reacts on this stimulation in any manner (output signal). Now we want to describe this system to predict the reaction of the system to an action. In the systems and control technology the relation of an input and an output function, and therefore, the system can be described by a transfer function (see Figure 4).

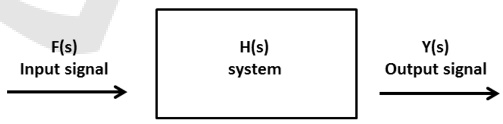

Figure 4: Relation between the output function $Y(s)$, the transfer function $H(s)$ and $F(s)$ the input function.

The mathematical relation is given by:

$$Y(s) = H(s)F(s), \tag{4}$$

where $Y(s)$ is the output function, $H(s)$ the transfer function and $F(s)$ the input function. If assuming, that the landing and recovery phase of a vertical jump (Figure 2 (e)) is an oscillating system, the transfer function is described by a second-order delay element (PT2-element).

$$H(s) = \frac{Y(s)}{F(s)} = \frac{a}{cs^2 + bs + 1} \tag{5}$$

Considering the general second-order system of an oscillator $H(s)$ can be described by

$$H(s) = \frac{K\omega_0}{s^2 + 2D\omega_0 s + \omega_0^2}, \tag{6}$$

where ω_0 is the natural frequency, K the DC gain of the system and D the damping ratio. It is assumed, that the error resulting from the use of the time-continuous model is small, because of a relatively high sampling ratio of 200 Hz. The natural frequency determines how fast the system oscillates during the response. The damping ratio determines how much the system oscillates as the response decays toward a steady state. These parameters can be deduced from equation 6, after transferring it in the form of equation 5:

$$H(s) = \frac{k}{(\frac{s^2}{\omega_0^2} + \frac{2Ds}{\omega_0} + 1)} \tag{7}$$

$$\omega_0 = \sqrt{\frac{1}{c}} \quad \text{and} \quad D = \frac{b\omega_0}{2} \tag{8}$$

Therefore, the natural frequency and the damping ratio can characterize the landing phase, the absorption of the impact, and the restoring of the balance and stability. As we will see in the next section, these parameters might be an alternative possibility to characterize the balance ability, the muscle strength, and allow conclusions to postural stabilization and neuromuscular control.

5 MODEL-BASED APPROACH

We propose the use of the oscillation behavior as an alternative approach to drawing conclusions about muscle strength, balance ability, postural stability, and neuromuscular control instead of using the DPSI, TTS, COP or ROM. The advantage of the model-based approach of the oscillation behavior (during landing and recovery phase) over existing amplification-based methods might be its potentially lower dependability on sample rates, and trial lengths.

In detail, we aim to model (as schematically illustrated in Figure 5) the human's lower extremities as a spring that oscillates during the landing and recovery phase. During free fall the spring is slack and will be compressed at the impact on the floor and the landing phase and depresses during the recovery phase to the steady state in one or more oscillations.

From a physical point of view, this system can be described by

$$F = -kx, \tag{9}$$

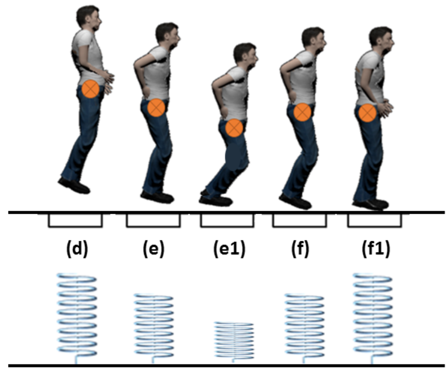

Figure 5: Comparing of the human's lower extremities with a spring during the landing and the recovery phase of a jump. The spring will be compressed during the landing (e-e1) and depresses during the recovery phase (f-f1) to the steady state in one or more oscillations.

with the force F, the displacement x and the spring constant k. The frequency can be estimated with

$$\omega_0 = \sqrt{k/m}. \tag{10}$$

This equation shows that the frequency correlates to the spring constant k.

In our model, the spring is characterized by the spring constant. Consequently, if comparing the spring with the muscles of the humans' lower extremities, the spring constant characterizes the stiffness of the muscles in a first approximation. Therefore, the natural frequency ω_0 of our system describes the ability to absorb the impact at the landing and characterizes the muscles of the lower extremities.

The damping ratio D indicates the influence within or upon an oscillatory system that has the effect of reducing its oscillations and might also be a relevant parameter for the characterization of the balance ability and the postural stability.

6 EVALUATION

6.1 Study Design

Each of the 20 considered healthy older adults of our study (12 subjects are female (60%) and 8 male (40%)) has performed three sequential CMJs with a rest of 1 min between the jumps to avoid signs of fatigue. Further characteristics of the subjects are listed in Table 2. The group covers a typical range of age, weight, and height for the group of pre-frail elderlies. The test procedures were approved by the local ethics committee (ethical vote: Hannover Medical School No. 6948) and conducted in accordance with the Declaration of Helsinki.

Table 2: Population characteristics of our study with the minimum (min.), maximum (max.), mean values and standard deviation (SD).

n=20	min.	max.	mean	SD
age [years]	71	82	75.1	3.02
weight [kg]	51.6	97.25	74.87	12.49
height [cm]	154.1	189.1	167.25	9.92

The jumps have been performed on an AMTI AccuPower ground reaction force plate, which is specified for jumping and power analyses and is an accepted gold standard. Figure 3 shows the coordinate orientation and the dimensions of the plate. The sampling rate amounts $200\,Hz$. The AccuPower sensitivity is based on a $8900\,N$ full-scale F_z capacity and a 12 bit internal AD ($\pm 2048\,bit$) or about 4.3 N/bit.

The transfer functions and FPE (as describing the transfer functions fit) have been estimated for each (of the 3x20) performed CMJs with Mathworks' MATLAB (version R2015a) using the System Identification Toolbox (version 9.2).

Per subject, the jump with the best fit estimation was taken into account in our analyses, whereby fits below 70% are rejected. Thereby, the function for the fit corresponds to the form of equation 5. An impulse function with the height of the impact force at landing will be assumed for the input signal. Figure 6 shows a typical input function.

Figure 7 shows a characteristic phase of landing and recovery and is assumed as an output function. To take the force in all directions into account, the absolute values for each axis are summed up. In the step of reprocessing the means are removed. In accordance with the theoretical considerations of Section 4, the number of poles is set to 2 and the number of zeros to 0 for the model of the transfer function, to describe an oscillating system (see equation 6). A discrete-time spectrum with T=0.005 s is chosen, because of the sampling rate of 200 Hz of the force plate.

6.2 Natural Frequency and Damping Ratio

The natural frequency ω_0 and damping ratio D were determined as described in section 4. The results for the natural frequency are in the range of about $1\,Hz$. Due to the literature, we expected higher frequencies (Wakeling et al., 2001). The analyze of the poles of the poles of the determined transfer functions show that there are only two real poles, which indicate a non-oscillating function instead of complex pole pairs, which lead to exponentially modulated oscillations. Consequently, the calculated parameter (called

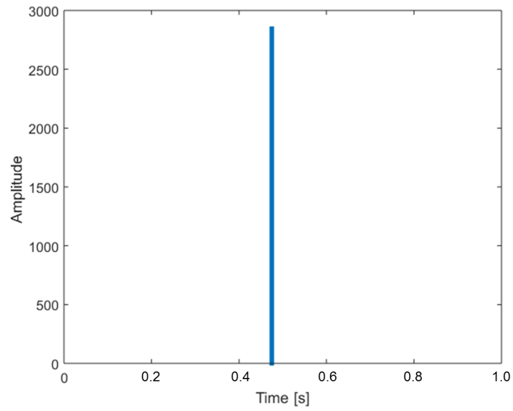

Figure 6: Typical impulse function as an input function.

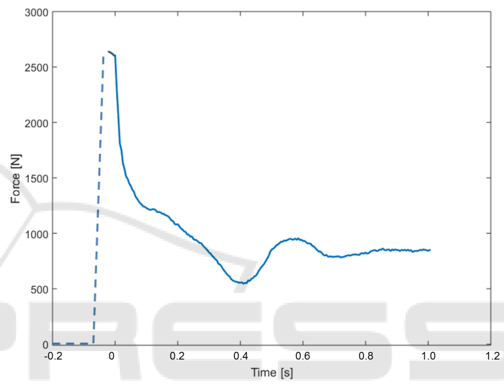

Figure 7: Typical phase of landing and recovery as an output function. The dashed line indicates the previous progress of the force during the flight phase.

w) is not equivalent to the natural frequency ω_0 and describes rather a damping factor. Nevertheless, the parameter w seems to be significant: Figure 8 shows the parameter w in relation to the jump height h. A linear regression results in

$$d = -0.0013 \cdot h + 1 \quad (11)$$

The relation shows an inverse relationship between these parameters. Therefore, the parameter w decreases with an increasing jump height.

Figure 9 shows the relation between jumping height and the calculated parameter d for the damping ratio D. Within the study group, the parameter d's effect size is small, since varying only in a small range. As an approximative assumption, the damping is linearized. Hence, an inverse relationship between the jumping height and the damping ratio was also recognized. The relation is not significant, due to the small effect size. It needs further investigations via data of both larger cohorts and heterogeneous groups, and additionally if the parameter d is equivalent to the damping ratio D.

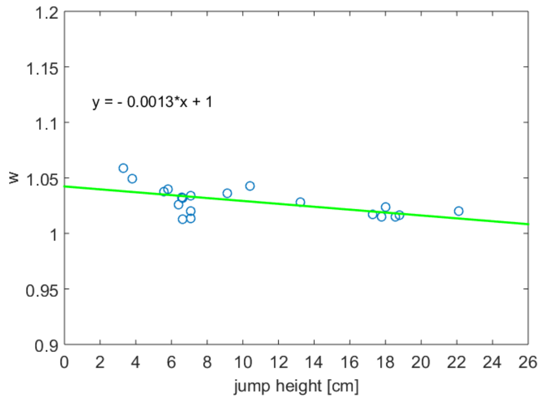

Figure 8: The parameter w is shown in relation to the jump height h. There is an inverse relationship between this parameter and the jump height.

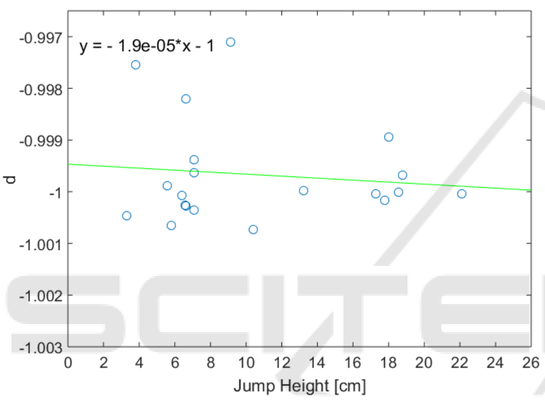

Figure 9: The estimated parameter d is shown in relation to the jump height h. The variation of this parameter lies only in a small range.

6.3 Major Factors of Influence

Due to the fact, that several factors can influence the parameters w and d, we analyzed the influence of the age, the sex, the body weight, the body height, w and d as a function of the jump height.

Therefore we used a generalized regression model. Due to the left skewed distribution of jump height (see Figure (10)) we estimated a gamma-distribution. Considering the jump height as clearly observable element, we performed a stepwise model selection by the Akaike information criterion (AIC).

The results of the regression analyses are listed in Table 3. As mentioned in Section 6.2 the parameter d varies only in a small range and is not significant. Also, the body height has no influence on the jumping height. But we can see a small influence of the body weight and an influence of the sex and the parameter w.

Table 3 shows the changes of the jump height for a one unit change of the listed parameters. For example, the sex has an influence of about $7\,cm$. The change of 0.01 unit of w results in a mean change of 1.5 cm of jump height.

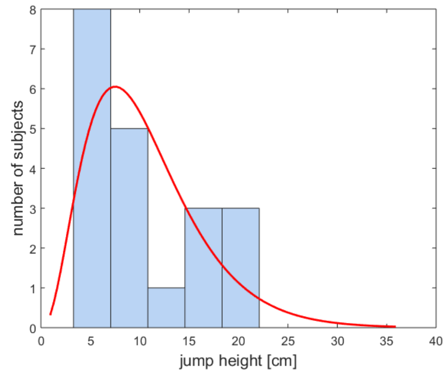

Figure 10: Distribution of the jump height and the estimated gamma-distribution (red line).

Table 3: Influence of weight, sex, and w depending on the jump height.

	Estimation	SD	$p(>\|t\|)$
w	-152.20	43.02	<0.01
weight	-0.073	0.04	0.08
sex	-7.34	1.95	<0.01

Figure 11 illustrates the differences between male and female subjects in jump height as a function of w.

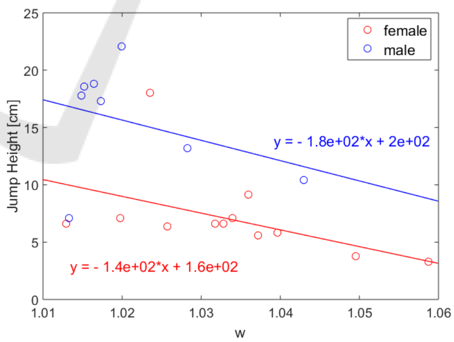

Figure 11: Comparison of the jump height as a function of w of male and female subjects.

Considering equation 10 in Section 5 and the simplified comparison of a human as a spring, the natural frequency increases with increasing spring constant and therefore increasing stiffness. On the assumption that the jump height h corresponds with the muscle strength S, there seems to be a relationship between jump height, the paramter w, muscle strength, and spring constant k:

$$h \sim S \sim \frac{1}{w} \sim \frac{1}{\omega_0} \sim \frac{1}{k}. \quad (12)$$

Thus, our model explains the inverse relation between jump height and the parameter w.

In order to analyze the postural stability, the dynamic postural stability index (DPSI) was determined in accordance with the approach of (Wikstrom et al., 2005b) by equation 13.

$$DPSI = \sqrt{\frac{\sum(0-F_x)^2 + \sum(0-F_y)^2 + \sum(m-F_z)^2}{n}}, \quad (13)$$

where F_x, F_y, F_z are the forces in anteroposterior (x), mediolateral (y) and vertical(z) direction, m the body weight, and n the number of data points. Figure 12 shows the resulting distribution of the DPSI over the considered CMJs of our study population. Next to the normal distribution of the DPSI values and the distribution-range (in relation to other age-related groups such as in (Wikstrom et al., 2005b; Meardon et al., 2016)) confirms the suitability of the DPSI for the considered group.

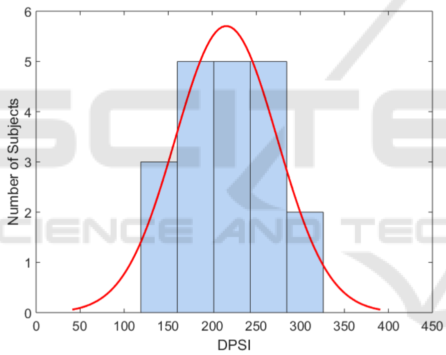

Figure 12: Distribution of the DPSI over our study population with a normal distribution fit.

In order to investigate the relation between w and postural stability (represented by the DPSI), we also analyzed the influence of the factors age, body weight, and w as a function of the DPSI. Table 4 lists the results.

Table 4: Influence of weight, sex and w depending on the DPSI.

	Estimation	SD	$p(>\|t\|)$
w	-1244.73	791.75	0.14
weight	2.25	0.816	0.01
age	7.71	3.37	0.04

We found an influence of these factors on the DPSI: For example, a change of 0.01 units in w results in a change of 12 of the DPSI-value. A change in the age of 10 years causes a change of 77 of the value of DPSI. While a significant correlation of the weight and age with the DPSI was found, the strong influence of w (indicated by an estimate of 1244.73, which is 161 times stronger than for age, the next strongest estimate) could not yet be statistically ensured due to the low number of subjects (n=20). Consequently, we are looking forward to confirming the expected significant correlation in an upcoming analysis with a larger study group.

7 CONCLUSION

While the CMJ is well-suited to measure muscle power and strength within a single assessment item, common traditional amplification-based balance measures for CMJ (such as TTS or CoP) have been shown to be significantly influenced by measurement settings including trial length, sample rate, and filter settings. Thus, a reliable alternative approach to detect balance ability for vertical jumps is required.

As an alternative approach, we propose to model the human body during the landing after a jump from a Systems and Control Technology perspective. Therefore, we used an impulse function of the maximum force at the impact on the ground as input function and the landing and recovery phase of a jump as output function. This phase is characterized by the balance and muscle strength of a subject.

Since the landing and recovery phase of vertical jumps can be represented by an oscillating system, the transfer function is described by a second-order delay element (PT2-element), where the natural frequency determines the systems oscillation frequency, and the damping ratio determines the system oscillation intensity as the response decays towards a steady state.

In an initial prospective study with 20 elderly probands, we could not observe the expected oscillating behavior in the phase of landing. Nevertheless, an inverse relationship between the calculated parameter w and the jump height and an inverse relation to the muscle strength could be determined. We could identify an influence of sex, weight, and w on the jump height.

Furthermore, a potential correlation between DPSI (as a common standard-index for balance) and w was seen but could not be clearly clarified due to the limited group size. Thus, we will investigate these effects in a further larger study.

Moreover, we considered in our model only one dynamic mode (one modal mass and one frequency). Using two-dimensional models of the musculoskeletal system (Blache and Monteil, 2013) or even an anatomically realistic three-dimensional musculoskeletal model (Farahani et al., 2016) could open perspectives for more robust models.

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